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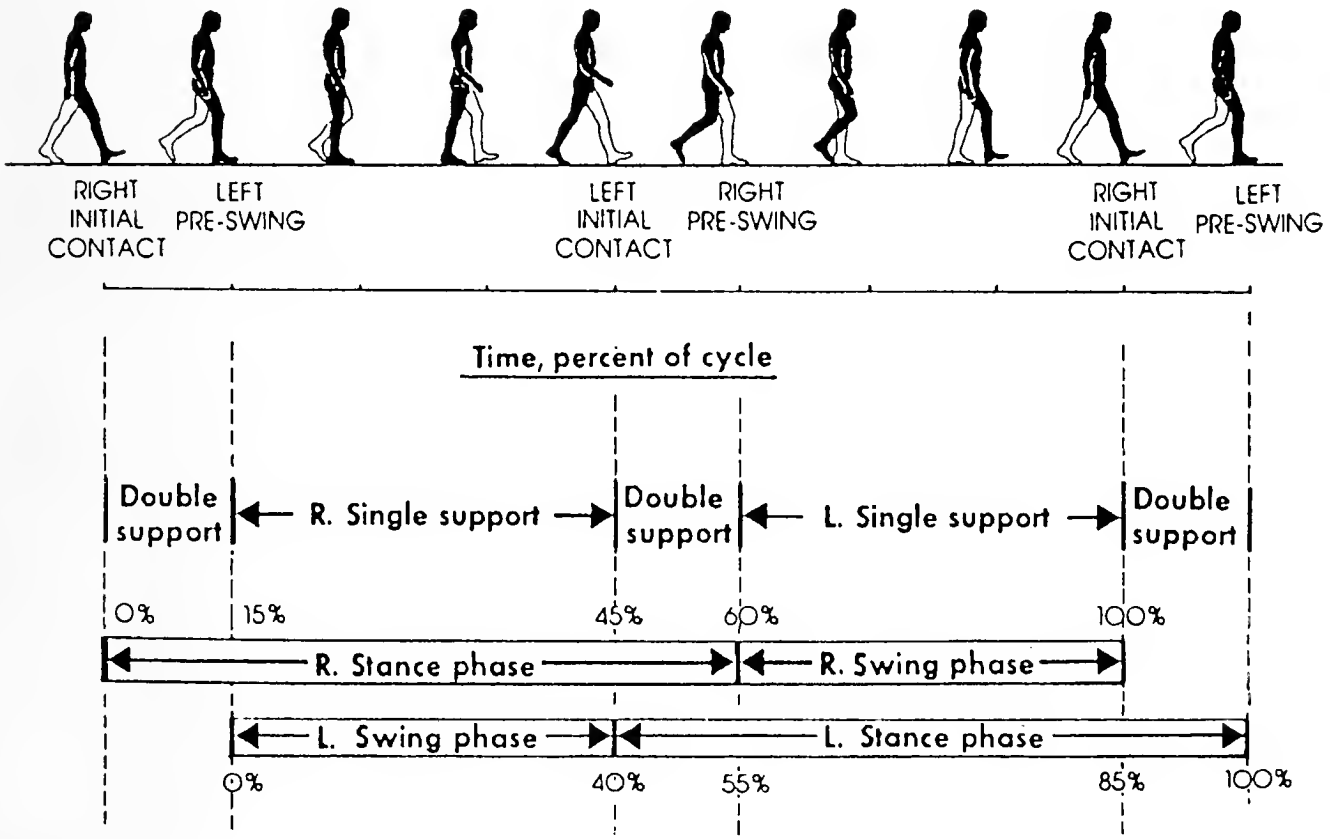
Gait Analysis in the Science of Rehabilitation

Joel A. DeLisa, M.D.



Department of Veterans Affairs - Veterans Health Administration

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Joel A. DeLisa, M.D.

Department of Veterans Affairs
Veterans Health Administration
Rehabilitation Research and Development Service
Scientific and Technical Publications Section

ON THE COVER

The cover design diagrammatically represents the time dimensions of the normal walking cycle, adapted from a figure that appeared on page 26 of a chapter by Verne T. Inman, MD, PhD, et al., in *Human Walking*, edited by Rose and Gamble, published by Williams & Wilkins, Baltimore, MD; 1981, and used with permission. Cover production: Frank Vanni.

Department of Veterans Affairs

Gait Analysis in the Science of Rehabilitation

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CONTENTS

Gait Analysis in the Science of Rehabilitation
Monograph 002, 1998

Joel A. De Lisa, MD

Before You Read This Monograph

This monograph, *Gait Analysis in the Science of Rehabilitation,* neither attempts to cover all instrumented gait analysis systems nor is to be considered a text on gait analysis. Other excellent source texts have been written and are referenced at the end of most chapters. It does, however, offer a solid foundation in understanding the availability and importance of instrumented gait analysis systems and how they work.

The intent of the chapter authors is to inform clinicians of the options open to them in their diagnosis and evaluation of pathological gait problems and what they might prescribe to improve or correct these deviations from the norm. This text has been designed so as to help clinicians find a way to reach an in-depth level of gait analysis in a clinical environment that may enable them to prescribe a quicker corrective solution, be it via an orthosis, a surgical procedure, or medication, thereby resulting in a speedier rehabilitation, discharge, and re-entry into independent living, which is the ultimate goal of both the clinician and the client.

Tamara T. Sowell, Editor

Joel A. DeLisa, MD, MS
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JOEL A. DELISA, MD, MS, Professor and Chairman of the Department of Physical Medicine and Rehabilitation, University of Medicine and Dentistry-New Jersey Medical School-New Jersey Medical School (UMDNJ); Senior Vice President and Chief Medical Officer of Kessler Rehabilitation Corporation; and President, Kessler Medical Rehabilitation Research and Education Corporation, is an experienced senior investigator, administrator, and teacher of medical rehabilitation. He received his M.D. from the University of Washington School of Medicine in 1968 and was Associate Professor at that university from 1981 to 1987. He has published 101 articles and 31 chapters and books. He is the Co-Director of the Physical Medicine and Rehabilitation (PM&R) Residency Training Program at UMDNJ-NJMS and maintains the integration of the research and clinical aspects of training, as well as a supportive institutional environment, for research. He has been certified by the American Board of Physical Medicine and Rehabilitation for 23 years and has written extensively in the field. The third edition of his textbook entitled *Rehabilitation Medicine: Principles and Practice* was released in July 1998. Dr. DeLisa is an active member of numerous professional organizations and has had wide-ranging experience on committees such as the Education and Examination Committees for the American Association of Electromyography and Electrodiagnosis (AAEE) and the Continuing Education Committee of the American Academy of Physical Medicine and Rehabilitation (AAPM&R).

Dr. DeLisa has been the Chairman of the Ad Hoc Committee on the Resident Matching Program of the Association of Academic Physiatrists (AAP) and a member of the AAPM&R/AAP Research Committee. He is past President of the AAEE, past President of the AAP, and past President of the American Paraplegia Society. He was an associate written examiner for the American Board of Physical Medicine and Rehabilitation (ABPM&R), is one of its 14 directors, and is its chairman. He is on the executive committee of the American Board of Medical Specialties and was a member of the Scientific Advisory Board of the Paralyzed Veterans of America and the National Multiple Sclerosis Society. He was on the Editorial Board of *Muscle and Nerve* and is currently a member of the following Editorial Boards: the *American Journal of Physical Medicine and Rehabilitation,* the *Journal of Spinal Cord Medicine,* and the *European Journal of Physical Medicine and Rehabilitation.* Among his many honors are The Special Advancement Award, in 1985, from the Seattle VAMC, for the development of the Spinal Cord Injury Program; The prestigious Gold Key Award, in 1991, from the American Congress of Rehabilitation Medicine; the Excellence Award, in 1995, from the American Paraplegia Society; the Outstanding Service Award from the Association of Academic Physiatrists; and the Charles L. Brown New Jersey Medical School Alumni Award in 1996. Dr. DeLisa was elected faculty member Alpha Omega Alpha in 1994.

From 1974 to 1987, Dr. DeLisa was affiliated with the Department of Veterans Affairs (then called the Veterans Administration) in the following capacities: from September 1974 to June 1975 as consultant in Rehabilitation Medicine at the American Veterans Hospital, Tacoma, WA; from June 8, 1975 to June 1982 as Assistant Chief of Rehabilitation Medicine Service, VA Medical Center, Seattle, WA; from October 7, 1979 to June 1982 as Associate Chief of Staff for Education, VA Medical Center, Seattle; and from August 20, 1984 to June 15, 1987 as Chief of the Spinal Cord Injury Service, also at the Seattle VAMC. He has served as liaison to the VA from 1993 to the present, as a member of the Special Medical Advisory (SMAG) to the Department of Veterans Affairs from 1992 to the present, and he has served on the VA congressional mandated Prosthetics and Assistive Devices Committee from 1991 to the present.

On November 6, 1998, Dr. DeLisa is scheduled to be the guest lecturer at the 31st annual Walter J. Zeiter Luncheon and Lecture, AAPM&R, which will be held at the Washington State Convention and Trade Center. This annual lectureship is awarded to a physiatrist who shares Dr. Zeiter's qualities of statesmanship, scholarship, executive leadership, and warm friendship. It is sponsored by the Physical Medicine and Rehabilitation Education and Research Fund (PM&R-ERF), AAPM&R.

EDITORIAL

by Joel A. DeLisa, MD, MS

I had the pleasure of doing my residency in physiatry under Justus Lehmann, MD, Professor, Department of Rehabilitation Medicine, at the University of Washington in Seattle, and, for my first five years on their faculty, taught functional anatomy to undergraduate physical therapy, occupational therapy, and prosthetic/orthotic students. Functional anatomy included the teaching of kinesiology and biomechanics, which became predominant as I taught both normal and abnormal gait to these students.

Locomotion is the process by which we move from one position to another. This process is a continuum from standing to walking to running and involves starting, stopping, changing directions, and altering speed. Most mammals are quadrupedal, but man is bipedal. The mastering of erect bipedal locomotion appears to be a learned activity and thus, each of us displays peculiarities that are superimposed on the basic pattern of bipedal locomotion. Therefore, on analyzing human gait, one should explain the similarities as well as the dissimilarities and then describe how these variations may represent an impairment.

In putting together this gait analysis monograph, I divided it into four sections: 1) clinical observation; 2) review of the instrumental gait analysis systems; 3) the value of information resulting from instrumented gait analysis from the perspective of a physiatrist, an orthopedic surgeon, and a physical therapist; and 4) discussion of future trends for gait laboratories. The authors were selected as experts from multiple rehabilitation specialties to give the readers an understanding of how gait analysis can be used to evaluate a person's walking abilities to maximize function and maintain or improve quality of life.

It is my belief that instrumented gait analysis systems offer objective evaluation of the effectiveness of the various rehabilitation treatments that are aimed at improving gait disabilities. Current recognized uses are in the gait patterns of persons with spastic paralysis to evaluate various orthopedic procedures such as tendon transfers, tenotomies, and rhizotomies pre- and postoperatively. In adults with other neurologic disorders and who exhibit spastic gait, quantitative analysis offers us objective data to evaluate therapeutic modalities and treatments such as strengthening and stretching exercises, biofeedback, functional electrical stimulation, various orthoses, and nerve or intramuscular neurologic blocks. One of the limits to its widespread use has been the limited reproducibility and usefulness of the data. Improvement in the computer technology, as well as the simplification of the systems, allows a much faster acquisition of kinematic data and analysis.

Clinicians will demand outcome studies with respect to who should be referred for these quantitative studies. Standardization of data collection and reporting procedures need to be implemented so that the cost-effectiveness relative to functional outcome can be established. Evidence-based practice is becoming a key driving force in medicine. This approach needs to be embedded in the determination of effective quantitative gait laboratories.

It is my hope that physicians and therapists in practice, as well as students at various levels of training, will find this monograph to be a user-friendly, valuable teaching tool.

I wish to thank the Department of Veterans Affairs Rehabilitation Research and Development Service for giving me this opportunity to organize and edit this monograph. I also want to thank Philip Melchiorre, MD, Assistant Professor of Physical Medicine and Rehabilitation, University Hospital, New Jersey Medical School, who was good enough to add his critique to the chapters.

D. Casey Kerrigan, MD, MS
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D. CASEY KERRIGAN, MD, MS, is Associate Professor and Director of Research at Harvard Medical School's Department of Physical Medicine and Rehabilitation (PM&R) and Founder and Director of the Center for Rehabilitation Science at Spaulding Rehabilitation Hospital, Boston, MA. She received her medical degree from Harvard Medical School and her residency training in PM&R from Cedars-Sinai Medical Center, West Los Angeles VA Medical Center, Children's Hospital of Los Angeles, Rancho Los Amigos Medical Center, and the University of California at Los Angeles, where she also received a Master's degree in kinesiology. She developed and directs the clinical and research Gait Laboratory at Spaulding Rehabilitation Hospital. Dr. Kerrigan also created the initial curriculum for the PM&R Residency Program at Harvard Medical School and served as its program director from 1994 to 1995. She was appointed a member of the committee to establish the Department of Energy/National Institutes of Health Lower Limb Prosthetics Project, where she served from 1996 to 1997. She is a founding member of the American Gait Laboratory Accreditation Board, which was established in 1995.

Among Dr. Kerrigan's many awards are the Ralph Goldman Intern of the Year Award from the West Los Angeles VA Medical Center (1988); the Outstanding Service Award for Resident Physician Council from the American Academy of PM&R (1992); the Young Academician Award from the Association of Academic Physiatrists (1996); and the First Special Recognition Award for Outstanding Teaching in the Harvard Medical School PM&R Residency Program (1996). She has published extensively in the area of gait and holds several foundation grants and a Clinical Investigator Award from the National Center for Medical Rehabilitation Research at the National Institutes of Health. Dr. Kerrigan is the recipient of grants from several foundations, the National Institutes of Health, and the Department of Veterans Affairs, all related to the subject of gait.

INTRODUCTION/PROLOGUE

by D. Casey Kerrigan, MD, MS

Modern-day quantitative gait analysis, including kinematic or joint motion measurement, kinetic or joint torque assessment, and dynamic electromyographic (EMG) recording, is one of the few, if not the only, measurement systems that quantify functional limitation, along with impairment and disability. Clearly, quantitative gait analysis allows an objective evaluation of the effectiveness of various rehabilitation treatments aimed at improving gait disability. For instance, quantitative gait analysis, as a functional assessment tool, has been used to show the benefits of various orthopedic surgical procedures and rhizotomy techniques in persons with neurological impairment. It has also been used to assess the biomechanical effects of bracing, prosthetic components, and other rehabilitative modalities.

However, functional assessment, or outcome measurement, is but one small role that quantitative gait analysis can play in the science of rehabilitation. If we expand the definition of gait analysis to include *interpreting* the significance of quantitative gait data, then the most promising aspect of gait analysis is that ultimately we will understand the complex relationships between impairment, functional limitation, and gait disability. An understanding of these relationships should vastly improve our rehabilitation treatment strategies.

The use of quantitative gait analysis in the rehabilitation setting has increased only recently. Gait analysis methodology has been around for over 100 years; however, work to improve gait analysis technology and repeatability has occurred only over the past 10 years. Often in the past, the technical details of gait analysis made *clinical* gait analysis extremely cumbersome and time-consuming. Two major factors made the routine use of gait analysis impractical, particularly in individuals with poor walking ability. The first factor was the time and effort required for setting up and testing a subject and the associated burden to that person. Depending on a particular protocol, obtaining kinematics with video capture could require that the individual not use an assistive device or that he or she walk with his or her arms crossed. The apparatus attached to the person was frequently heavy and constricted joint movement. The second major limitation was the time and effort required to process and analyze the data. Also, unreliability of data acquisition and processing methods required an inordinate number of trials, burdening both the people being tested and the staff. Such limitations undoubtedly prohibited gait analysis on a regular clinical basis.

There is a growing acceptance of the clinical use of gait analysis in the rehabilitation setting. Fortunately, in recent years many technical difficulties have been overcome. Recent advances, such as improved computer processing and the development of passive as opposed to active marker systems, have enabled the faster acquisition of kinematic data without heavy encumbering attachments and wires trailing from the subject. Also, with improved computer integration and software, kinetic data are more automatically obtained from a combination of kinematic and force plate data. Although once impractical, a modern-day gait laboratory can now allow for routine assessment of gait in standard rehabilitation settings.

Currently, the most common clinical use of gait analysis is the assessment of spastic paretic gait. Analysis allows us to understand the dynamic implications of a particular impairment, such as spasticity or weakness, in a particular muscle group. For instance, in some individuals, spastic paretic stiff-legged gait, defined as reduced knee flexion during the swing period of the gait cycle, may be the result of quadricep spasticity. In fact, a standard, static evaluation may reveal spasticity in the quadriceps. A gait analysis, however, may or may not demonstrate inappropriate activity in the quadriceps during the critical phases of the gait cycle when the knee should be flexing in preparation for and during swing. In this way, gait

analysis allows us to determine the functional implication of an impairment. In some cases, by using gait analysis we may observe an impairment or functional limitation that is not at all appreciable with static evaluation. For instance, despite normal tone and the absence of spasticity in a particular muscle group such as the quadriceps, gait analysis may reveal inappropriate activity in that same muscle group.

With gait analysis, we have the potential to determine those impairments and functional limitations that probably contribute to the walking disability. Although a gait disability may be phenotypically similar from one individual to another, the impairments are typically distinct between individuals. Probably no two sets of quantitative gait data from two individuals are the same, no matter how visually similar their gait disability appears. Logically, the optimal treatment for a given individual will be the one that addresses the impairments and functional limitations that are most likely contributing to the walking disability. Gait analysis should provide this information, thereby allowing an effective rehabilitation management program. That gait analysis can help define the appropriateness of a number of rehabilitative modalities is probably its most important potential contribution to rehabilitation science.

By defining the causative impairments and functional limitations, a gait analysis can be used to focus and optimize rehabilitation treatment including the prescription, for instance, of specific strengthening or stretching exercises, EMG biofeedback, functional electrical stimulation, orthotics, or nerve or intramuscular neurolytic blocks. A quantitative gait evaluation may identify which muscles are firing appropriately and which are not, based on the kinematics, dynamic EMG, and kinetics. Gait analysis especially fills a void in upper motor neuron pathology, where traditional static evaluation measures are not effective in measuring either muscle strength or spasticity, at least from a functional standpoint. By providing information as to which muscle groups need strengthening (or electrical stimulation, or bracing), and which need relaxation (or stretching or intramuscular neurolysis), gait analysis can lead to a more optimal, methodical, and directed rehabilitation protocol.

By helping to pinpoint the causative impairments and functional limitations, gait analysis could be quite useful in optimizing experimental protocols involving a number of rehabilitation treatments. For instance, an experimental EMG biofeedback or functional electrical stimulation experimental protocol that was based on information obtained from quantitative gait analysis undoubtedly would be more likely to be successful than one that was not. Similarly, a program to test therapeutic modalities aimed at reducing spasticity would be more likely to be effective if it were based on information obtained from quantitative gait analysis. For example, a program to reduce tone in the quadriceps to improve stiff-legged gait would be functionally helpful only if gait analysis revealed that the quadriceps really were inappropriately active during gait. Gait analysis is thus potentially quite useful in optimizing, and thereby increasing, the likelihood of demonstrating the general effectiveness of a number of innovative, as well as standard, rehabilitation treatments. Of course, quantitative gait analysis, by providing objective functional assessment information, also can be helpful in assessing the outcomes of these specific rehabilitation programs.

The future of gait analysis in the science of rehabilitation is bright. Undoubtedly, its technology will continue to improve with new developments in computer vision, artificial intelligence, computational methods, and computer power. In addition, the models used to interpret gait analysis data will be refined and standardized. Perhaps the most exciting area of development is that of forward dynamic or robotic modeling, in which much work has already been done. Ultimately, we ought to be able to input kinetics that are measured with current gait analysis technology into a computerized robotic model. Inputting an individual's measured kinetics into the robotic model would result in a kinematic gait pattern that is similar to the individual's actual kinematic pattern. Then, we could predict the effect of changing the kinetic inputs; changing them in the robotic model would be the equivalent in the actual individual to some rehabilitation treatment that would alter an impairment, such as strengthening a particular muscle. We would be able to alter kinetic inputs in the model and observe the "would

be" kinematic pattern changes. The ability to model and predict the effect of a treatment would not only provide a better understanding of the mechanisms of various gait disabilities, it would allow each person more precise individualized rehabilitation prescriptions for treatment.

With improved measurement techniques, gait analysis will continue to provide us with a better understanding of biomechanical and neurophysiologic function, both normal and abnormal, which may transcend to other activities of daily living. The role of gait analysis in the science of rehabilitation is much larger than simply a functional assessment tool as it can help us determine the complex relationships between impairment, functional limitation and disability. By defining these relationships, we not only will be able to design more optimal studies of the general effectiveness of a number of current rehabilitation treatment strategies, but will also be in a better position to consider new rehabilitation treatment strategies.

SECTION ONE

Clinical Observation

by Gerard Malanga, MD and Joel A. DeLisa, MD

Dr. Malanga is Director of Spine and Occupational Medicine at Kessler Institute for Rehabilitation and Assistant Professor of Physical Medicine and Rehabilitation at the University of Medicine and Dentistry-New Jersey Medical School. Dr. DeLisa is Professor and Chairman of the Department of Physical Medicine and Rehabilitation at the University of Medicine and Dentistry-New Jersey Medical School and is the President of Kessler Medical Rehabilitation Research and Education Corporation.

INTRODUCTION

The ability to walk upright is a defining characteristic of man. Gait is the manner in which walking is performed and can be normal, antalgic, or unsteady. Gait analysis can be assessed by various techniques but is most commonly performed by clinical evaluation incorporating the individual's history, physical examination, and functional assessment. Gait abnormalities can be more precisely examined through the use of gait laboratories. These laboratories utilize surface EMG activity of muscles, force plates, and kinematic evaluation of the lower limbs. They are highly specialized units that assess various gait abnormalities from individuals with neuromuscular disorders to high-level athletics. While some clinical impressions have been shown to be incorrect by the use of gait lab technology, the clinical evaluation still remains the essential component in determining the etiology and the treatment plan for gait problems. A proper clinical evaluation should always precede any gait lab assessment.

Normal Gait

The determination of abnormal gait requires one to first have an understanding of the basic physiology and biomechanics of normal gait (1,2,3). The gait cycle is a time interval or sequence of motion occurring from heelstrike to heelstrike of the same foot. The gait cycle has been broadly divided into two phases: stance phase and swing phase. These phases can then be further subdivided and discussed in terms of percentage of each within the gait cycle. This is diagrammatically represented in **Figure 1,** by Verne T. Inman, MD, PhD.

The stance phase is 60 percent of the gait cycle and can be subdivided into double-leg and single-leg stance. In double-leg stance, both feet are in contact with the ground. At an average walking speed, it represents 10 percent of the entire gait cycle, but decreases with increased walking speed and ultimately disappears as one begins to run. At slower walking velocities the double-leg support times are greater. Single-leg stance comprises up to 40 percent of the normal gait cycle. The muscles that are active during the stance phase act to prevent buckling of the support limb. These include the tibialis anterior, the quadriceps, the hamstrings, the hip abductors, the gluteus maximus, and the erector spinae (1,4,5).

The swing phase is described when the limb is not weight bearing and represents 40 percent of a single gait cycle. It is subdivided into three phases: initial swing (acceleration), midswing, and terminal swing (deceleration). Acceleration occurs as the foot is lifted from the floor and, during this time, the swing leg is rapidly accelerated forward by hip and knee flexion along with ankle dorsiflexion. Midswing occurs when the accelerating limb is aligned with the stance limb. Terminal swing then occurs as the decelerating leg prepares for contact with the floor and is controlled by the hamstring muscles.

Determinants of Gait and Energy Conservation

During gait, three main events occur in which energy is consumed. This includes controlling forward

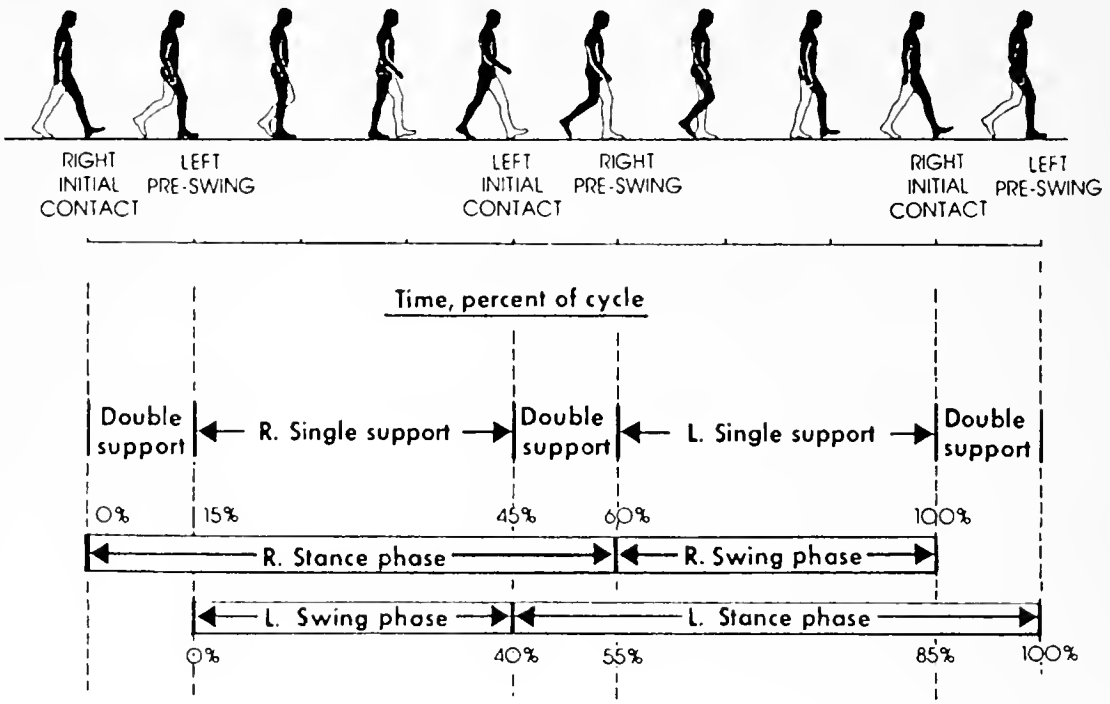

Figure 1.
Time dimensions of the gait cycle. (Reprinted, with permission, from a chapter by V.T. Inman et al., which appeared on page 26 of *Human Walking*, edited by Rose and Gamble and published by Williams & Wilkins, Baltimore, MD; 1981.)

movement during deceleration toward the end of swing phase, shock absorption at heelstrike, and propulsion during push off, when the center of gravity is propelled up and forward (6,7). Muscle activity used during the gait cycle is noted in **Table 1.**

A human's center of mass (COM) is located just anterior to the second sacral vertebra, midway between both hip joints. The least amount of energy is required when a body moves along a straight line, with the COM deviating neither up nor down, nor side to side. Such a straight line would be possible in normal gait if man's lower limbs terminated in wheels instead of feet. This obviously is not the case, thus, our COM deviates from the straight line in vertical and lateral sinusoidal displacements.

With respect to vertical displacement: the COM goes through rhythmic upward and downward motion as it moves forward. The highest point occurs at midstance, the lowest point occurs at time of double support. The average amount of vertical displacement in the adult male is approximately 5 cm.

With respect to lateral displacements: As weight is transferred from one leg to the other, there is shift of the pelvis to the weight-bearing side. The oscillation of the COM amounts to side-to-side displacement of approximately 5 cm. The lateral limits are reached at midstance.

In his classic article, Inman describes the components of gait (8). These are referred to as the six determinants of gait (**Table 2**). He describes several mechanical factors that help to flatten the arc in the vertical and horizontal (lateral) planes reducing displacement of the body's COM and thereby reducing the energy expenditure. The net effect is a smooth, sinusoidal translation of the COM through space along a path that requires the least amount of energy. Any pathology that increases the vertical distance between the high and low points, increases the energy cost of ambulation.

First determinant: pelvic rotation in the horizontal plane. This allows the swinging hip to move forward faster than the stance hip (1–3,8,9). Pelvic rotation

Table 1.
Primary muscular activity during the gait cycle.

Muscular Activity	Muscles	Period
Shock Absorbers	Quadriceps Dorsiflexors	Weight-Loading
Stabilizers	Gluteus Maximus, Medius, & Minimus Tensor Fascia Lata Erector Spinae	Stance-Phase
Foot Lift Off	Flexor Digitorum Longus Flexor Hallucis Longus Gastrocnemius Peroneus Longus and Brevis Soleus Tibialis Posterior	Weight-Unloading
Accelerators	Adductor Longus and Magnus Iliopsoas Sartorius	Weight-Unloading
Foot Controllers	Extensor Digitorum Longus Extensor Hallucis Longus Tibialis Anterior	Swing-Phase
Decelerators	Gracilis Semimembranosus Semitendinosus Biceps Femoris	Swing-Phase mid-swing to initial-contact

Table 2.
Determinants of Gait.

Determinant		COM Displacement	Effect
First	Pelvic Rotation About the vertical axis, alternating to the right and to the left relative to line of progression	Decreased 4° of each side from a total of 8° Reduces the drop in COM during double limb support	Energy conservation saves the COM drop at its lowest point 6/16 inch (elevates end or arc)
Second	Pelvic Tilt At horizontal axis at midstance	Reduces the peak of COM during single limb support	Energy conservation by shortening the pendulum of the leg (3/16 inch) at the high part of arc (depresses summit arc)
Third	Knee flexion in stance	High point of COM further reduced by knee flexion in midstance	Energy conservation by decreasing rise of arc (7/16 inch) by walking over a bent knee (depresses summit arc)
Fourth & Fifth	Foot and ankle mechanism	Combination of foot and ankle motion with knee motion smoothes the COM change in direction	Flattens and slightly reverses arc of translation (decreased 3/16 inch)
Sixth	Lateral displacement of pelvis	Must bring COM above support point to balance on one leg	Lateral displacement of the pelvis is largely abolished by the presence of the tibial-femoral angle. There is a side-to-side sway of 1.7 inch radius

COM = Center of Mass

occurs anteriorly on the swinging limb and posteriorly during midstance. It is maximal just before heelstrike with a total motion of pelvic rotation of 3–5° to each side. Pelvic rotation also produces a longer stride length for the same amount of hip flexion of the advancing leg and hip extension of the retreating leg. Thus, it allows for longer steps without changing the COM displacement significantly.

Second determinant: pelvic tilt in the frontal plane. As the pelvis on the swing leg is lowered, the hip abductors of the stance hip control pelvic tilt. During normal gait, the pelvis drops 4–5° away from the stance leg and toward the swing leg. This pelvic dip decreases horizontal displacement of the COM during single limb support.

Third determinant: knee flexion, which acts to decrease vertical displacement of the COM. This occurs during midstance, as knee flexion to approximately 15° occurs under the control of eccentric quadriceps contraction and remains flexed until the foot is flat on the ground. These first three determinants save one inch of vertical displacement with each stride.

Fourth and Fifth determinants: involve control of the knee-ankle-foot motion. This synchronized movement results in eccentric control of plantar flexion of the ankle and knee flexion, which occurs during the first portion of the stance phase. These factors help to avoid abrupt changes of the lowest portion of COM arc, producing a smooth, sinusoidal curve instead of an arched pattern.

Sixth determinant: lateral pelvic movement. This is the lateral sway or side-to-side oscillation that occurs with each step. This defines the motion of the COM in the horizontal plane. The shifting of the pelvis occurs over the supporting foot to provide stability during the stance phase. The extent of sway is determined by the base of support. Normal knee valgus between the femur and tibia helps to reduce the amount of pelvic shifting required for stability and allows the feet to be closer together during forward progression.

Murray et al. (10) determined parameters of gait in nondisabled men. They found that the mean duration of the gait cycle was 1.03 seconds. The steps per minute were 117 (90–120 steps) and the average comfortable walking speed was 2.8 miles per hour. The average stride length was 70–82 cm and the average stride width was 8 cm, with a foot angle of 6.7°. Ostrosky et al. compared gait characteristics in young and old subjects and found that older people demonstrate less knee extension and a shorter stride length compared with younger people (5).

The clinical evaluation of gait occurs within the context of a detailed history and physical examination. The history may reveal complaints of pain, weakness, or instability. In addition, it is important to know the individual's past medical history to be aware of underlying neurologic or musculoskeletal problems. The examination must include a detailed musculoskeletal and neurologic examination. It must address an evaluation of the person's muscle strength, joint range of motion, tone, and proprioception. The musculoskeletal examination should include, at a minimum, the joint above and below the area of complaint. The entire kinetic chain, which includes the spine and the upper limb, should be considered. The history and physical are helpful in focusing the differential diagnosis of the complaint. The observation and evaluation of gait can occur either before or after the physical examination and is included as a part of the physical examination. It is the authors' preference to evaluate the individual's gait after a detailed history and physical examination. **Figure 2** and **Table 3** summarize the main muscle actions and their timing during the gait cycle.

GAIT ANALYSIS

The observation of gait begins with a general assessment, noting symmetry and smoothness of movements of the various body parts (**Table 4**). The clinician should take note of the cadence (steps/minute), base width, stride length, arm swing, movement of the trunk, and rise of the body. The observer must then look at the individual segments of the kinetic chain as the subject ambulates, including the head, shoulders, arms, trunk, pelvis, hips, knees, ankles, and feet. Specifically, the clinician examines the head position, whether the shoulders are depressed, elevated, protracted, or retracted. The amount of arm swing can be categorized as normal, increased, or decreased. The trunk may have a forward or backward lurch or a list to the right or left. The pelvis may be hiked, level, dropped, or fixed. The hip may demonstrate increased extension, flexion, rotation, circumduction, or an adducted or abducted posture. The knee is observed for proper flexion, extension, and general stability in the various phases of the gait cycle. The ankle is examined for plantarflexion and dorsiflexion, as well as eversion and inversion.

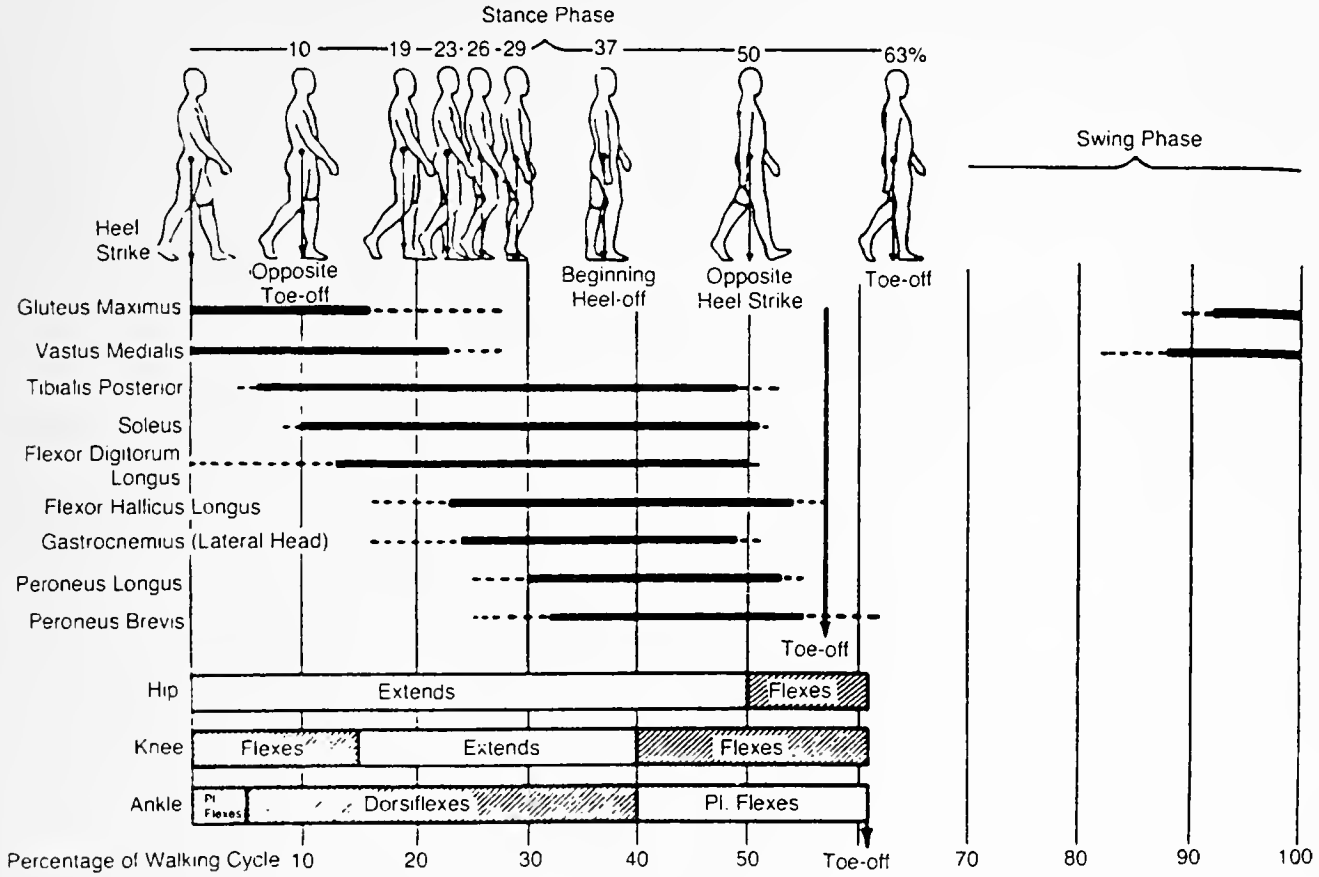

Figure 2.
On-off patterns of electromyographic activity of ankle plantar flexors. (Reprinted, with permission, from an article entitled "An electromyographic study of the plantar flexors of the ankle in normal walking on the level," by DH Sutherland, which appeared on page 66 of the *Journal of Bone and Joint Surgery Vol. 48A:1966.)*

Finally, the foot is observed for proper push off and excessive pronation and supination during weight bearing. If pain is experienced during walking, the subject should so indicate, so that its position in the gait cycle can be identified.

Because the entire gait cycle ends in a little over one second, a systematic and disciplined approach must be used to clinically evaluate a person's gait. Subjects should be viewed from the front, side, and behind, while they are wearing a minimal amount of clothing. The front view is helpful in viewing any deviations of the trunk or pelvis. One can also look for proper upper limb swing, which is usually opposite of the pelvis and lower limb. Upper limb swing helps to balance and smooth the forward progression of the body. The side view is helpful in examining exaggerations of spinal motions (e.g., hyperlordosis and hip motion). It is also best for observing the load response of the stance leg. One can look for ankle plantarflexion, knee flexion followed by ankle dorsiflexion, and hip and knee extension, as the leg progresses to midswing. The ankle should demonstrate proper plantarflexion at initial contact and then dorsiflexion in midstance through just before heel off. The posterior view is probably best for observing pelvic abduction or adduction in determining whether there is a Trendelenberg gait.

PATHOLOGIC GAIT

Pathologic gait patterns can be broadly divided into either neuromuscular or musculoskeletal etiologies

Table 3.
Main muscle function for unimpaired ambulation.

Muscle	Gait Cycle Function
Gastrocnemius and soleus	Midstance to heelstrike
Gluteus maximus	Heelstrike to midstance
Gluteus medius and minimus	Heelstrike to toe off
Hamstrings	Midswing to heelstrike
Iliopsoas and adductors	Toe off to midswing
Quadriceps	Heelstrike to midstance Toe off to midswing
Tibialis anterior and peroneals	Heelstrike to foot flat Toe off to heelstrike

Table 4.
Gait: Major points of observation.

1. Cadence a. Symmetrical b. Rhythmic	6. Pelvic a. Anterior or posterior tilt b. Hike c. Level
2. Pain a. Where b. When	7. Knee a. Flexion, extension b. Stability
3. Stride a. Even/uneven	8. Ankle a. Dorsiflexion b. Everison, inversion
4. Shoulders a. Dipping b. Elevated, depressed, protracted, retracted	9. Foot a. Heelstrike b. Push off
5. Trunk a. Fixed deviation b. Lurch	10. Base a. Stable/variable b. Wide/narrow

(3,5,11,12). Gait deviations may be a result of structural abnormalities of the bone, joints, or soft tissue. Limitations of lower limb joint mechanisms will usually be compensated by increased motion at the joints above and below (12). Other general causes of pathologic gait include neuromuscular and myopathic conditions or painful segments of the lower limb kinetic chain. Generally, as the efficiency of the gait pattern is reduced, the energy expenditure is increased.

Common Musculoskeletal Causes of Pathologic Gait

Hip Pathology

Osteoarthritis is the most common abnormality of the hip resulting in gait abnormalities. The first changes noted are diminished hip range of motion especially in internal rotation and flexion. This often results in exaggerated compensatory motion in the lumbar spine and the opposite unaffected hip. In severely restricted hip joints, there will be a reduction in hip flexion in the swing phase and in hip extension during the stance phase. These restrictions will be somewhat compensated by other joints (e.g., hip hiking on the unaffected side or "tiptoeing" on the affected side).

The antalgic gait is the most common pattern seen in individuals with a painful hip. This is characterized by avoidance of weight bearing on the affected side and a decrease in the stance phase on that limb in an attempt to unload the mechanical stresses on the painful hip joint. In addition, a trunk lurch toward the painful hip of the stance leg brings the COM over the joint and decreases the mechanical stress across the joint. This is done by dipping the shoulder on the affected side, elevating the opposite shoulder and shifting the pelvis over the stance leg during the stance phase of the gait cycle. During the swing phase, the hip is slightly flexed, externally rotated, and abducted in order to relax the joint capsule and ligaments to reduce joint tension. Heelstrike tends to be avoided in persons with a painful hip in order to prevent jarring and excess loading of the joint.

Knee Pathology

In general, a painful knee is maintained in slight flexion throughout the gait cycle. This is especially true if there is an intra-articular effusion, as slight flexion reduces the tension on the knee joint capsule (12). Compensation for knee flexion involves the avoidance of heelstrike and toe walking on the affected side. This type of antalgic gait may result from any painful condition of the knee joint including a meniscal tear, loose body, fracture, infection, or inflammatory synovitis.

Ligamentous instability of the knee can result in variable gait presentations depending on the ligament involved. The most common gait pattern seen in ligamentous laxity, hyperextension, or "recurvatum," is a result of a loss of muscular control of the knee secondary to various neuromuscular problems. In these cases, the knee must rely on the static stabilizers (i.e., the ligaments and joint capsule), which become stretched and lax over time. During the stance phase of the gait cycle, the knee hyperextends, which, over time, leads to degenerative changes of the knee joint.

Another abnormal gait pattern associated with instability of the knee ligaments is the varus thrust gait pattern seen in persons with injuries of the posterior-lateral corner of the knee. These injuries usually involve a combined injury to the posterior cruciate ligament, lateral collateral ligament, posterior joint capsule, and the popliteus tendon. The combined injuries, can lead to significant functional impairment requiring reconstructive surgery. The gait pattern seen in these people is characterized by varus thrust, which occurs at the knee during the stance phase of gait. They should be differentiated from isolated injuries of the lateral collateral or posterior cruciate ligaments, which generally have a good prognosis with nonoperative treatment.

The quadriceps avoidance gait occurs in those who have suffered an injury to their anterior cruciate ligament (ACL). The quadriceps muscle provides an anterior force to the tibia, which becomes a problem in someone with an ACL deficiency, as the tibia is prone to anterior subluxation. The person will attempt to decrease the load response phase on the affected limb by decreasing the stride length and avoiding knee flexion during the mid-portion of stance (13,14).

Knee-joint contractures will also lead to abnormal gait patterns. A flexion contraction of the knee will cause signs of a short leg limp. A flexion contracture of less than 30° becomes more pronounced with faster walking speeds, while contractures of more than 30° are apparent with normal walking speeds (12). The gait is characterized by toe walking on the affected side and a steppage gait or hip hiking on the unaffected side.

Foot and Ankle Pathology

Painful conditions of the foot and ankle from trauma, inflammatory disorders, degenerative arthritis, and so forth, will result in an antalgic gait pattern. There will be an attempt to limit weight bearing through the affected area. The stride length will be greatly shortened and normal heel-to-toe motion will be lost. If the problem involves the forefoot, the person will tend to avoid plantarflexion and toe off. If the problem involves the ankle or hindfoot, then the person will avoid heelstrike at initial contact and will ambulate with a tiptoeing gait on the affected side with compensations on the unaffected side.

People with ankle instability will have great difficulty with supporting body weight during initial contact on the stance leg. At contact, the unstable ankle will often buckle with a resultant antalgic gait limiting the load response phase on the affected side.

Joint contractures of the ankle are often seen after trauma, immobilization, and neurologic problems affecting the muscles of the ankle and foot. The most common contracture seen in clinical practice is contracture of the gastrocsoleus complex or "heel cord." A tight or contracted heel cord will result in a steppage type gait pattern. There will be a loss of normal heel contact and heel-to-toe motion, along with exaggerated hip and knee flexion during the swing phase in order to clear the toe. In long-standing contracture, hyperextension of the ipsilateral knee may occur as plantarflexion at the ankle causes an extension moment at the knee.

Problems of the hindfoot, particularly of the calcaneus, will produce a similar gait pattern (e.g., elimination of heelstrike and a promotion of toe contact during stance). These problems include calcaneal fractures, plantar fasciitis, stress fractures of the ankle or calcaneus, and so forth. An antalgic or avoidance gait with a decrease in the loading of the heel is the typical pattern. In contrast, problems of the forefoot (sprain, fracture, arthritis, metatarsalgia, etc.) will result in an antalgic gait, which minimizes loading on the forefoot by decreasing plantarflexion during the stance phase and push off. People with these problems will tend to increase loading to the heel and hindfoot, and shorten the time of forefoot loading.

Leg Length Discrepancy

Leg length discrepancy can be the result of various factors affecting any segment of the kinetic chain including scoliosis and contracture of the hip, knee, and ankle, and is termed a "relative" leg length discrepancy. A true leg length discrepancy is the result of asymmetry in length of the pelvis, femur, or tibia. In either case, a leg length discrepancy can result in pelvic obliquity with a drop of the pelvis, decreased hip and knee flexion, ankle plantarflexion, and/or hyperpronation, which all occur ipsilateral to the shortened side. It is important to determine the etiology of the leg length discrepancy and to properly treat the underlying cause rather than treating all leg length discrepancy with a heel lift. In leg length discrepancy of less than 1.27 cm during the entire stance phase, one sees dipping of the shoulder on the affected side and a compensatory pelvic drop. There is an apparent elevation of the shoulder on the opposite (swing side) and an exaggerated flexion of the hip, knee and ankle on the ipsilateral side. For shortening more than 3.81 cm, he or she will walk on

tiptoes on the shortened limb during the stance phase with full knee extension.

Neurologic Causes of Abnormal Gait

Any dysfunction of the central nervous system, spinal cord, peripheral nerve(s), or muscle(s) can result in an abnormal gait (2,3,12). It is important to know the segmental innervation of the trunk and lower limbs to evaluate for abnormal gait patterns, particularly the peripheral nerve innervation of each muscle and region. In addition, neurologic injury may result in changes in motor tone and control. The more common disease problems leading to pathologic gait will be reviewed.

Hemiplegic Gait

Cerebrovascular injuries commonly result in various gait abnormalities, the most common of which is the hemiplegic gait. It is characterized by abnormal arm swing with the arm carried in adduction with flexion at the shoulder, elbow, wrist, and fingers. Also, in many people, there is an extensor synergy of the affected lower limb, consisting of extension, adduction, and internal rotation at the hip, extension at the knee, and plantarflexion and inversion of the ankle and foot. This synergy pattern is often initiated by weight bearing over the involved limb and can be useful in supporting the subject. The hemiplegic gait tends to be quite slow with a decrease in step length and an increase in the stance phase with circumduction to allow toe clearance. Compensatory changes include hip hiking from lack of knee flexion of the stance leg, a decreased lateral shift over the affected side, a lack of heelstrike secondary to the plantarflexion of the ankle, and recurvatum of the affected knee. The extension moment at the knee is created by the plantar flexion moment occurring at the ankle. Swing phase is characterized by an absent or markedly reduced knee flexion due to quadriceps spasticity. The flexor synergy gait occurs less commonly and consists of hip flexion, abduction and external rotation, knee flexion, and ankle dorsiflexion. This synergy pattern does not allow the person to stand, thereby eliminating ambulation potential.

Spastic Gait

A spastic gait can develop from an insult to the central nervous system that affects motor tone, particularly of the lower limbs. This can result in "scissoring" of the lower limbs from over-activity of the hip adductors and a narrow, crossing base. There is associated tiptoeing to maintain balance and great effort is exerted to swing the legs forward, all of which create an unsteady fatiguing gait. In addition, isolated muscles or muscle groups may develop increased tone and spasticity. For example, spasticity of the tibialis posterior, a powerful plantarflexor and inverter of the foot, causes significant changes in gait during both the stance and swing phases. During stance phase, the initial contact will occur on the lateral aspect of the foot and plantar flexion at the ankle results in an extension moment at the knee. Plantarflexion will also result in a relative lengthening of the limb; often causing dragging of the toes and requiring increased hip and knee flexion.

Parkinsonian Gait

Parkinson's disease results from lesions of the basal ganglia affecting motor control and function bilaterally. It is characterized by a paucity of movement of the facial, trunk, and upper and lower limb muscles. This results in a gait that is slow and shuffling with short rapid steps described as being festinating. The trunk is flexed forward and the person may have difficulties with stops and turns, appearing to chase after his or her COM (12). Joint motion is reduced due to rigidity and there is usually little or no arm swing to help in balancing the individual, with falls being a common result.

Ataxic Gait

Injury to the cerebellum or its pathways may disrupt the normal coordination and precision of motor function. The gait of these individuals will be unsteady and associated with a broad standing base and a lurching or staggering of the trunk and lower limbs. Movements are uncoordinated and appear exaggerated (4). Leg placement will be variable and reproducibility is lost. An ataxic gait may also be seen in persons with sensory deficits of the lower limb. In these people, the base is wide, and there may be slapping of their feet as they hit the ground. In addition, these individuals will tend to look at their feet due to the lack of proprioceptive feedback and, therefore, have more problems at night or in the dark.

Isolated Motor Weakness Gait Problems

Gluteus Maximus (Lurch) Gait

The gluteus maximus, a major hip extensor and stabilizer of the trunk, prevents the trunk from falling forward as the COM moves forward at heelstrike. In weakness of the gluteus maximus, the hip is supported by the ligament of Bigelow, which becomes taut in

hyperextension. The individual will throw the hip backward with a "lurch" using abdominal and paraspinal muscle activation just after heelstrike on the affected side. The backward trunk lurch persists throughout stance to maintain the gravitational force line behind the hip axis locking the hip in extension. There is an apparent forward protrusion of the affected hip due to the exaggerated trunk motion and the person may also hold the shoulders backward to keep the center of gravity behind the hip joint. The hamstring muscles will often compensate for isolated gluteus maximus weakness resulting in a near normal gait pattern; however, these muscles are often affected together (e.g., in S-1 radiculopathy).

Gluteus Medius (Trendelenberg) Gait, Uncompensated or Compensated

In uncompensated gluteus medius weakness, there is a drop of the pelvis more than the usual 5° on the unaffected side beginning with heelstrike on the affected side and continuing until heelstrike on the unaffected side. There is also a lateral protrusion of the affected hip. In compensated gluteus medius gait due to severe or total paralysis of the hip abductors, the pelvic drop appears to be less as the subject laterally bends the trunk over the hip and drops the shoulder on the affected side. This serves to keep the center of gravity over the hip, which decreases the muscle force required to stabilize the pelvis. With both compensated and uncompensated gait, because the affected leg becomes functionally longer, there is an increase in hip and knee flexion and ankle dorsiflexion. This steppage gait allows for toe clearance.

Hip Flexor Weakness

Hip flexors are the major accelerators in the swing phase of gait. Weakness of the hip flexors results in a limp starting during the stance phase of gait at push off persisting throughout the swing phase of the affected side. The subject will demonstrate a trunk lurch backward and toward the unaffected side from push off to midswing. This results in locking of the hip joint on its ligaments, with further extension of the trunk as a unit from push off to midswing carrying the affected leg forward. The inertia generated from trunk and hip activity carries the limb into flexion. The stride thus becomes shortened on the affected side.

Quadriceps Weakness

Weakness of the quadriceps is most apparent during heelstrike through the stance phase of gait. However, the limp affects all phases of the gait cycle. The affected knee must be locked in hyperextension at or preceding heelstrike by compensatory activity of the gluteus maximus extending the femur and the soleus, which extends the tibia. Extension at the femur results in flexion of the trunk and an extension moment at the knee. Some people place their hand on their thigh at heelstrike and stance to assist the knee into this extended position. With rapid walking, the affected leg lags during swing phase resulting in excessive heel rise. Repetitive hyperextension of the knee results in stretching of the ligaments and capsule of the knee and resultant recurvatum of the knee during the stance phase.

Ankle Dorsiflexor Weakness (Drop Foot, Slap Foot, or Steppage Gait)

With mild weakness, the gait abnormality will be noted at heelstrike and results in loss of plantarflexion control. Heelstrike to foot-flat phase occurs rapidly and the foot may slap at heelstrike, as eccentric control of the dorsiflexors is decreased. In severe weakness or paralysis, the foot will fall into plantarflexion during swing phase, presenting as a footdrop. Heelstrike is absent and the person comes down with the toes first or with the entire foot. This will cause a relative lengthening of the limb, compensated for by exaggerated hip and knee flexion to allow for toe clearance (steppage gait).

Gastrocsoleus Weakness

Gastrocsoleus weakness results in loss of ankle dorsiflexion control. Heel off will be delayed and the push off phase will be decreased. This results in a lag of forward movement of the pelvis on the unaffected side at the time of heelstrike and on the affected side during push off. There will be a shortening of the stride on the unaffected side due to the delay of forward movement of the ipsilateral hip. Altered ground reactive forces lead to a flexion moment behind the knee, which can result in knee buckling (10).

CONCLUSION

The ability to walk upright is a key functional activity which, when performed abnormally, impacts adversely on activities of daily living. The clinical evaluation of gait abnormalities, performed in conjunction with a thorough history and physical examination, is an important undertaking. These gait abnormalities

result from various neuromusculoskeletal disorders and can often be detected during the screening evaluation. Making the proper diagnosis is important in allowing for appropriate rehabilitation and/or orthotic strategies. Occasionally, for managing complicated spasticity or for determining surgical correction, a formal gait laboratory evaluation may be necessary.

REFERENCES

1. Lehmann JF, de Lateur BJ, Price R. Biomechanics of normal gait. Phys Med Rehabil Clin North Am 1992;3:125–38.
2. Magee DJ. Orthopedic physical assessment, 3rd ed. Philadelphia: W.B. Saunders; 1997.
3. Schneck C. Normal and abnormal gait. Ninth Annual Review Course in Physical Medicine and Rehabilitation, 1998, March 6–15, West Orange, NJ. West Orange; NJ: Kessler Institute for Rehabilitation. p. 2–1 to 2–9.
4. Lehmann JF, Condon SM, de Lateur BJ, Smith JC. Gait abnormalities in tibial nerve paralysis: a biomechanical study. Arch Phys Med Rehabil 1985;66:80–5.
5. Ostrosky KM, VanSwearingen JM, Burdett RG, Gee Z. A comparison of gait characteristics in young and old subjects. Phys Ther 1994;74:637–46.
6. Klopsteg PE, Wilson PD, editors. Human limbs and their substitutes. New York: McGraw-Hill; 1954.
7. Inman VT. Conservation of energy in ambulation. Arch Phys Med Rehabil 1967;47:484–8.
8. Inman VT. Human locomotion. Can Med Assoc J 1966;94:1047–54.
9. Saunders JB, Inman VT, Eberhart HD. The major determinants in normal and pathological gait. J Bone Joint Surg 1953;35A:543–58.
10. Murray MP, Drought AB, Kory RC. Walking patterns of normal men. J Bone Joint Surg 1964;46:335–60.
11. Gyory AN, Chao EYS, Stauffer RN. Functional evaluation of normal and pathologic knees during gait. Arch Phys Med Rehabil 1976;57:571–7.
12. Lehmann JF, de Lateur BJ, Price R. Biomechanics of abnormal gait. Phys Med Rehabil Clin North Am 1992;3:125–38.
13. Andriacchi TP. Dynamics of pathological motion: applied to the anterior cruciate deficient knee. J Biomech 1990;23 (Suppl 1):99–105.
14. Berchuck M, Andriacchi TP, Bach BR, Reider B. Gait adaptations by patients who have a deficient anterior cruciate ligament. J Bone Joint Surg Am 1990;72A:871–7.

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SECTION TWO

Instrumented Gait Analysis Systems

by Ernest L. Bontrager, MS

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INTRODUCTION

The measurement of human gait has come a long way in the past 40 years. Modern gait analysis started with the work of Inman and Eberhart (1–3) in the 1950s and became a useful clinical tool through the pioneering efforts of Perry (4–7) and Sutherland (8,9). These pioneers were able to show the clinical value of relating muscle function to joint motion and phases of the gait cycle, which resulted in surgical procedures to improve the gait of those suffering from spastic paralysis and other neuromuscular disorders.

That these early researchers obtained clinically useful results is all the more amazing when one considers the basic instrumentation available to them. Most of the instruments were pieced together from various sources (10) and/or developed "in house." Joint motion was measured from custom-made electrogoniometers or laboriously digitized by hand from motion picture films (9). Raw electromyography (EMG) was recorded on analog tape recorders and displayed with footswitch timing information on "Visicorder" strip charts. Hand measurement of footswitch timing from these records was used to calculate temporal gait parameters. A roomfull of strip chart albums at the Pathokinesiology Laboratory of Rancho Los Amigos Medical Center testifies that, with proper dedication and effort, a lot can be done with less than optimal tools.

The computer age has brought with it a much brighter picture for today's clinician who wishes to perform clinical gait analyses. From relatively inexpensive devices to very costly systems, the necessary tools are readily available to equip a modern gait lab. The large number of vendors provides many options from which to choose when selecting gait instrumentation (**Table 1**). Unfortunately, with all these suppliers, confusion can arise as to how to spend gait instrumentation dollars.

The purpose of this article is to provide information on the types of gait instrumentation that are commercially available and give some criteria for selecting the appropriate instrumentation. Also included are unique and/or key features of each manufacturer's products. This will not be a "Consumer Report" type of article, as I have not used or tested all the instruments reported here. Instead, this report is based on my understanding of gait instrumentation in general, the material provided to me by the manufacturers, and personal communication with other engineers and users of these systems.

PICTURE VIDEO

Techniques have been developed to enable a trained observer to make critical judgments about an individual's gait, by viewing a video recording of the person walking (11). Hence, one of the simplest pieces of gait instrumentation also is one of the most useful. A picture video system allows the clinician to record a person's gait prior to applying any instrumentation (EMG electrodes, footswitches, motion markers, and so forth) that might alter the gait patterns. It provides visual documentation of what occurred during the instrumented tests and is the only way of resolving differences when the recorded footswitches or motion data do not correspond to the clinician's visual image of the subject.

Table 1.
Gait instrumentation manufacturers by type.

Manufacturer	Picture Video	Temporal Gait			Foot Pressure		Motion		Force		EMG				
											Electrodes		Acquisition		Analysis
		Foot Switch	Mats	Other	Mats	Insoles	Goni	Video	Plates	Load Cells	Surface	Wire	Wireless	Cable	
AMTI									X	X					
Ariel								X							
Bertec									X	X					
Biometrics Ltd.							X								
B & L Engineering	X	X		DL							A				X
Bortec											A			W, FO	
BTS	X	X		X				X			A		X	FO	X
Charnwood Dynamics				X				X							
CIR Systems			X												
Delsys											A				
EQ Inc.			X												
IOMED											A				
tVM											P				
Kistler									X	X					
Konigsberg													X	FO	
Market-USA				X			X			FS			DL		X
Motion Analysis				X				X							
Motion Lab Systems		X									A			W	X
Musgrave Systems					X										
Nicolet Biomedical											P	X			
Noraxon													X	W	X
Northern Digital								X							
Novel Electronics					X	X									
Oxford Metrics				X				X							
Peak Performance	X			X				X							
Qualisys				X				X							
Sensor Medics											P				
Tekscan				X	X	X									

FS = Force Sandals A = Active P = Passive DL = Data Logger W = Wire FO = Fiber Optic Goni = Goniometer

A basic video system consists of a VCR, one or two video cameras, a character generator, a video mixer, and a TV monitor. The video mixer combines the images from two cameras so that an anterior/posterior (A/P) and lateral view can be observed simultaneously. Some users find the two views confusing and prefer to combine a simultaneous record of EMG and/or footswitch data on oscilloscopes with a single view of the person walking. The character generator enables one to overlay text (e.g., name, date) on the video image. Three manufacturers provide picture video systems (**Table 2**) that were designed to be used with specific

Table 2.
Picture video system features.

Manufacturer	Recorder	Storage Media	Used With	Software Controlled?
B & L Engineering	VCR	Tape	Vicon Motion System	Yes (1)
BTS	VCR	Tape	BTS ELICLINIC	Yes
Peak Performance	DVR	Disk	Peak Motus Motion System	Yes

DVR = Digital Video Recorder
(1) VCR must be manually operated if used without the Vicon.

motion systems. The clinician should check with the manufacturer if he or she wants to use it with another system or as a stand-alone system.

TEMPORAL GAIT MEASUREMENTS

Since gait is repetitive in nature, temporal gait measurement systems provide the clinician with a valuable analytical tool in gait analysis by quantifying the timing of critical events in the cycle. Cadence, gait cycle duration, stance and swing times, single limb support, and initial and terminal double limb support are typical parameters measured. By making the measurements over a defined walking distance, average velocity and stride length also can be defined. Measuring only velocity and single limb support can reveal a great deal about an individual's functional ability to ambulate. As that person gets weaker, has painful joints, or feels unstable, velocity will decrease and less time will be spent in single limb support on the affected side.

Footswitches

Footswitches are a convenient and inexpensive way of obtaining temporal gait measurements. There are two basic types, compression closing and force sensitive resistor (FSR) switches, usually configured as thin insoles, which can be placed between the foot and shoe or taped to the bottom of a bare foot.

Compression closing switches consist of a sandwich of thin pieces of brass shim stock separated by a compressible (nonconducting) foam rubber insole (**Figure 1**). In the contact areas, conductive rubber cylinders are inserted into holes in the insole. When pressure is applied, the insole compresses and the conductive rubber cylinders contact the pieces of brass on each side of the insole, closing an electrical circuit. This sandwich is held together with duct tape and is typically about 4-mm thick.

The FSR switches consist of two thin layers of flexible plastic, with printed circuits on the inner surfaces, separated by a thin layer of double-sided adhesive. Holes in the adhesive create contact areas. As pressure is applied, carbon on one surface contacts a metal pattern on the other surface, creating a resistive electrical circuit (**Figure 2**). As more pressure is

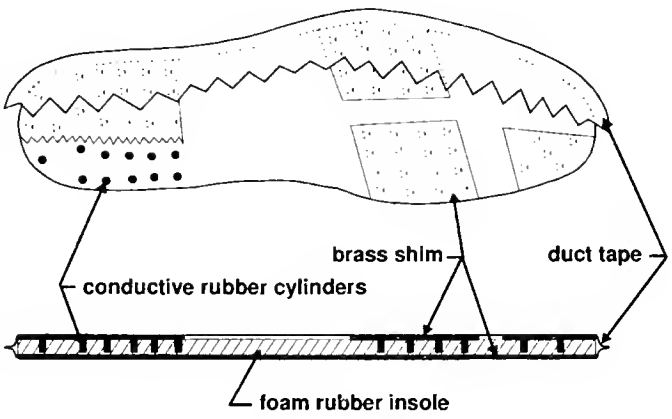

Figure 1.
A typical compression closing footswitch (not to scale). For clarity, the duct tape, which holds the "sandwich" together, is only shown along the lateral edge on the top view. The cross-section view shows the conductive rubber cylinders and brass shim, as well as the two duct tape layers.

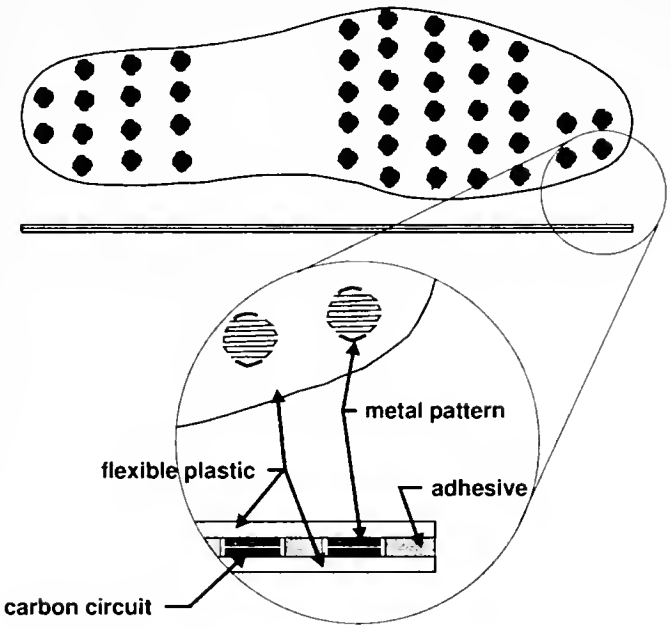

Figure 2.
A force sensitive resistor (FSR) footswitch (not to scale). An enlarged view of a portion of the toe section shows more detail of the flexible plastic layers with the printed circuit contact areas on the inner surfaces. Pressure causes carbon on one surface to contact a metal pattern on the other surface, creating a resistive electrical circuit. As more pressure is applied, the resistance drops to a level that is detected as a switch closure. The interconnecting printed circuit traces are not shown.

Figure 3.
A person walking down the "GaitRite" gait mat, free of any encumbering equipment. The computer at the right displays the foot/floor contact patterns as the switches in the mat close due to foot pressure. The temporal and spatial gait parameters are calculated and displayed for printing and storing in the database. (Photograph used with permission.)

applied, the resistance drops. The associated circuitry triggers, at a predefined resistance value, indicating a switch closure.

Footswitches typically have contact areas in the heel, first and fifth metatarsal, and great toe areas (**Table 3**). Some facilities use discrete switches taped to critical areas under the foot rather than an insole, which incorporates the switches into a single module. The advantage of discrete switches is that different sizes are not required to fit a large range of foot sizes. The disadvantage is in getting reliable data because of difficulty in consistently placing the switches at the proper locations under the foot.

Typical footswitch activation delay times, as compared with force plate data, are from about 1 to 2 percent of the gait cycle. For a nonimpaired person walking with a 1-second gait cycle, this is a delay of about 10 to 20 msec at both initial and terminal contact. Some footswitch software compensate for this delay.

Some facilities obtain temporal gait data from their video motion systems, identifying foot-floor contact from the motion marker trajectories. A disadvantage of this technique is that the temporal resolution is restricted to the frame rate of the video system (20 msec for a 50 Hz frame rate).

In addition to the footswitches, B & L Engineering (Tustin, CA) manufactures the Footswitch Stride Analyzer, a computer-based instrument that computes all of the temporal gait parameters based on footswitch data averaged over a measured distance. This system also provides a graphic representation of the foot-floor contact patterns (12). The user wears a small battery-powered microcomputer recorder unit (data logger), which stores up to four runs of data. An optical link is used to download the data to a PC for calculation and printing of the results. A light-sensitive switch worn on the user's upper arm triggers the recorder when he or she passes special triggering lights set up at the beginning and end of the measured walkway.

Gait Mats

Gait mats are relatively new systems that provide both temporal and spatial gait parameters. These mats consist of a long strip of walking surface, such as carpet, into which is embedded an array of switches running across and along the length of the mat (**Figure 3**). As a person walks down the mat, the switches close under the feet, enabling the computer to calculate the timing of each switch closure. Since the geometry of the mat is known, the spatial parameters of gait can be

Table 3.
Footswitch features.

Manufacturer	Type	Areas of Contact	Thickness (mm)	Size
B & L Engineering	Compression Closing (Insole)	Heel 5th Met 1st Met Great Toe	3	All Standard Male & Female Sizes (1)
BTS	FSR (Insole)	Heel Lat Foot Med Foot	2	140,191, 216 & 267 mm long
Motion Lab Systems	FSR (Discrete)	User Selected	0.5	18 & 28 mm dia.

FSR = Force Sensitive Resistor Met = Metatarsal Lat = Lateral Med = Medial (1) Can be custom made to user selected sizes.

calculated. Besides step length measurements, the advantages of these systems are the elimination of any gait encumbering attachments, low cost, and portability. The major disadvantages are the spatial resolution due to the finite size of the switches and the temporal resolution due to limitations in the scan rate. Both systems (**Table 4**) provide an extensive database and have provisions for editing the raw data file if desired.

FOOT PRESSURE

Capacitive and FSR transducers are the two basic types in use today for plantar pressure measurement. The capacitive transducers consist of two capacitor plates separated by a compressible rubber dielectric material. As pressure is applied, the capacitor plates are pushed closer together resulting in increased capacitance, which is calibrated in units of pressure. The FSR transducers are fabricated in a manner similar to that described for the FSR footswitches. As pressure is applied to the transducer, the electrical resistance decreases, indicating an increase in pressure. The accuracy of these systems is dependent on the ability to reliably calibrate them, as the transducers tend to be nonlinear. Pneumatic pressure bladder calibration systems generally are used. Since the area of the transducers is known, the applied force can be calculated by adding up the force computed from each active sensor at a given point in time. These systems are valuable, because they provide a method of quickly determining the areas of high pressure on the plantar surface of the foot, areas that may be subject to tissue breakdown. Two types of systems, mats and insole devices, are available commercially.

Pressure Mats

A pressure mat is placed in the center of the walkway and used much like a force plate, with the subject stepping on it as he or she walks down the walkway. It provides a quick and easy way of obtaining a plantar pressure picture, as nothing needs to be attached to the individual. However, if the effects of shoe insoles or various orthoses are to be evaluated, an insole pressure-measuring device must be used. All three pressure mat systems listed (**Table 5**) are factory calibrated and have software that includes color pressure pictures, gait lines, force and pressure versus time, force and pressure/time integrals, and masks for detailed analysis of selected areas of the foot.

Pressure Insoles

Pressure insoles were designed to provide the same kind of data available from pressure mats, with the added advantage of in-shoe measurement and multiple cycles. Dynamic measurement of footwear and orthoses is possible with these insoles. One can quickly and easily compare the plantar pressure distribution with different shoe inserts and/or orthoses. With special care, barefoot data can also be obtained by lightly taping the

Table 4.
Gait mat features.

Manufacturer	Type	Active Area	Thickness (mm)	Switch Spacing (mm)	Temporal Resolution (msec)	Special Features/Considerations
CIR Systems (GaitRite)	Portable (1)	61 cm x 3.66 m	4	12.7	11	Can handle walking aid patterns Computes FAP score (3)
EQ. Inc. (GaitMat)	Transportable (2)	61 cm x 4.17 m	32	15	10	Needs 32 mm thick runways at each end for pre & post walk area (4)

(1) Can be rolled up and carried in a convenient plastic golf case.
(2) Folds into four 99 x 41 x 1 cm pieces that fit in a storage case.
(3) The Functional Ambulation Performance (FAP) score is a single numerical representation of a person's gait, based upon temporal and spatial gait data as well as the person's physical measurements (10*, 13*).
(4) Manufacturer does not provide runways.
*Reference numbers.

Table 5.
Pressure mat features.

Manufacturer	Sensor Type	Size (mm)	No. of Sensors	Sensor Density (per cm^2)	Sample Rate (Hz)	Calibration	Special Features
Musgrave Systems (Musgrave Footprint)	FSR	194 x 394 x 38	2,048	2.7	55.6	Dynamic Force	Double Plate System Available
Novel Electronics (EMED)	Capacitive	225 x 445 x 20	2,016	2	70	Static Pressure Bladder	Podometry Software Provided
Tekscan (F-Mat)	FSR	320 x 470 x 6	2,128	1.4	120	Static Force	Real Time Display

FSR = Force Sensitive Resistor

transducer to the bottom of the foot. The insole must be protected from possible damage and the clinician must insure that the floor/insole interface does not create a slipping hazard for the wearer. Both the Pedar and F-Scan systems (**Table 6**) incorporate the software developed for the pressure mats manufactured by their respective companies.

MOTION

Since walking involves cyclical movement patterns at multiple joints, it is important to measure these kinematic patterns as a basis for interpreting other gait data (EMG, force, stride characteristics). The kinematic measurements (which also include limb segment velocities and accelerations) are necessary for the determination of joint moments and forces (kinetics).

Two basic types of motion measurement systems are in use today: electrogoniometers and video motion systems. Although other techniques exist—hand digitized film (9), strobe light photography (10,14,15), and electromagnetic—they have either been replaced by newer technologies or never caught on as a clinically useful tool.

Electrogoniometers

Electrogoniometers are electro-mechanical devices that span a joint to be measured, with attachments to the proximal and distal limb segments (**Figure 4**). These

devices provide an output voltage proportional to the angular change between the two attachment surfaces. They operate on the assumption that the attachment surfaces move with (track) the midline of the limb segment onto which they are attached and, thereby, measure the actual angular change at the joint.

The two major advantages of these devices are low cost and ease of use. As is the case with all gait instrumentation, care must be exercised in applying them to the individual. The tracking assumption is reasonable for lean individuals, but the more "fleshy" and/or muscular the person being tested, the less likely the true angular change will be recorded due to skin and muscle movement. When considering these devices for gait, their accuracy should be carefully evaluated by testing them on individuals of various statures. The person should move through a known range of motion (i.e., 90°) while the goniometer output is being recorded. This will give a general idea of the kinds of errors the clinician might encounter.

A number of different potentiometric goniometers have been developed for gait. They were designed to cause a potentiometer shaft to rotate proportionally to the joint angle being measured. Various designs were incorporated to allow for the polycentric joint axis at the knee. One of these designs, the double parallelogram goniometer, has been used with considerable success at the Pathokinesiology Laboratory, Rancho Los Amigos Medical Center. The double parallelogram linkage allowed translation of the attachment cuffs to occur without creating a change in the potentiometer output. This device is not commercially available.

Biometrics Limited (Penny & Giles, Inc., Santa Monica, CA) has developed strain gauge goniometers that are light, flexible, and easy to use. They consist of a small diameter, tightly coiled, flexible spring with plastic endblocks on each end (**Figure 4**). The strain gauge mechanism housed inside the spring, changes electrical resistance proportionally to the change in angle between the longitudinal axes of the endblocks. One endblock is telescopic, compensating for changes in the distance between the endblocks as the limb moves. The endblocks are attached to the limb segments with double-sided adhesive tape. These devices are biaxial, enabling one to simultaneously measure sagittal and frontal plane motions. They come in various sizes, to accommodate different joints, and have a very large functional measuring range (greater than 180°). This company also makes similarly designed "torsiometers" for measuring axial rotations. For instrumentation, they provide a data logger, which stores the data for later downloading to a PC via a serial port (software is available). They also manufacture a four-channel amplifier that consists of a small portable body-worn unit and a larger tabletop base unit, for connecting to a strip chart recorder or computer A/D converter.

Infotronic (Market-USA, Inc., Severna Park, MD) sells a goniometer system that incorporates the Penny & Giles transducers described above. Their system has a data logger that stores the angle data on memory cards (see EMG Acquisition Systems, below). The data can later be downloaded to a PC. The software enables the user to plot angle/time and angle/angle diagrams.

Video Motion

Video systems utilize one or more video cameras to track bright markers placed at various locations on the person being tested. The markers are either infrared (IR) light-emitting diodes (LEDs) for active marker systems or solid shapes covered with retroreflective tape for passive marker systems. The systems keep track of the horizontal and vertical coordinates of each marker from each camera. In three-dimensional (3D) systems, the computer software computes 3D coordinates for each marker based upon the 2D data from two or more

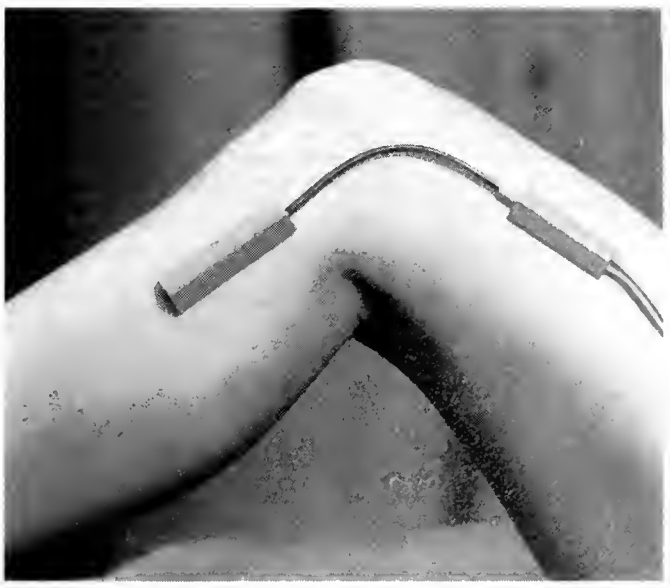

Figure 4.
A Penny & Giles strain gauge electrogoniometer applied at the knee. The strain gauge in the small spring measures the angle between the plastic endblocks that are attached to the leg with double-sided adhesive tape.

Table 6.
Pressure insole features.

Manufacturer	Sensor Type	Thickness (mm)	Sizes	No. of Sensors	Max. Sample Rate (frames/sec)	Calibration	Software
Novel Electronics (Pedar)	Capacitive	2.5	12 standard sizes (1)	99 - adult 84 - child	58	Static Pressure Bladder	Pedar Step Analysis (5) & EMED
Tekscan (F-Scan)	FSR	0.2	scissor trimmed (2)	960 max. (3)	100	Body Weight or Pressure Bladder (4)	F-Mat & Langer EDG for temporal gait

FSR = Force Sensitive Resistor
(1) Standard sizes range from 160 to 300 mm in length. Custom sizes available.
(2) Can be trimmed from men's 14 to child's size 3.
(3) Variable depending on trim size.
(4) Accuracy: Body Weight: 10%; Pressure Bladder, 3 to 5%.
(5) Step Analysis provides 2 & 3D pressure pictures, step timing and pressure and force as a % of the cycle.

cameras and the known location of all cameras. In practice, more than two cameras are needed, as markers become obscured from camera views because of arm swings, walking aids, and/or patient rotation.

If only one camera is used (2D), the assumption is that all motion is occurring in a plane perpendicular to the camera axis. This is seldom the case and any marker movement outside this plane will be distorted. As a result, 2D systems are not recommended for gait and should only be used in very controlled situations.

It should be pointed out that just because a system computes the 3D coordinates of each marker, it does not mean, *a priori*, that 3D kinematics will be produced. To obtain true 3D motions, each body segment must be defined by at least three markers (which create a plane passing through the segment), joint centers must be defined, and Euler angles computed. Knee and ankle joint centers are either determined from width measurements or medial markers used only during a calibration (''quiet standing'') test. Most commercially available systems provide software that attempts to determine true 3D kinematics (**Table 7**). Prior to purchasing a system, the buyer should ensure that he or she understands the assumptions in the kinematic modeling and their impact on the results. For example, most systems utilize a common marker on the lateral femoral epicondyle for both the thigh and shank segments. This hinge joint approximation at the knee may introduce errors with large flexion angles. The calculated hip joint center is often used in place of one of the thigh markers, a technique that can introduce errors in thigh motion. Some systems do not measure inversion/eversion at the foot due to the difficulty of placing three closely spaced markers on the foot.

Kinetics software computes the net joint moments, forces, and powers based upon the kinematics, ground reaction forces, and anthropometric data. Most provide kinetics in all three planes. As with the kinematics, one should be comfortable with the models used, and the way segment mass and moments of inertia are approximated.

All of the systems provide the capability of acquiring at least 16 channels of analog data simultaneously with the motion data (**Table 7**). Most compute temporal gait parameters measured from bilateral motion data if footswitches are not used. Most gait motion data are collected at a frame rate of 50 or 60 Hz, so temporal gait measurements utilizing the motion data will have a minimum time resolution of 20 and 16.7 msec, respectively, as compared with 2 msec or less for typical footswitch systems. The camera's field of view limits the number of strides available. Unlike footswitch systems, however, step length can be obtained from motion data.

Two important factors to consider for any clinical application are ease of use (which includes processing speed) and accuracy. Ehara et al., conducted a performance comparison (accuracy, marker noise, and processing speed) of nine video 3D motion systems (16). In addition to three systems available only in Japan and a

Table 7.
Video motion system features.

Manufacturer	System Type	Input Device	Comp. O.S.	Analog Channels		Calib.?	Temporal Gait?	Marker ID	Kinematics				Kinetics
				No.	Rate (Hz)				3D	Foot Inv/Ev	Knee & Ankle Joint Centers		Clinical Software
											Patient Calib.?	Measure-ments?	
Ariel (APAS)	Passive	VCR's	W	32	2K	Yes	No	SA (2)	Yes	Yes	No	Yes	Sag, Fr Tr
BTS (Elite)	Passive	Video Camera	W, WNT DOS	64	1K	Yes	Yes	SA	Yes	(4)	Optional	Yes	Sag, Fr, Tr
Charnwood Dyn. (CODA mpx30)	Active	Scanner Camera	W, WNT	24	2K	No (1)	Yes	A	Yes	Yes	No	Yes	Sag, Fr Tr
Motion Analysis (ExpertVision)	Passive	Video Camera	WNT, Un, SG	64	5K	Yes	Yes	SA (3)	Yes	Yes (5)	Yes (6)	Yes	Sag, Fr, Tr
Northern Digital (Optotrak)	Active	Video Camera	DOS	16	4K	No (1)	No	A	Gait Software Not Provided				
Oxford Metrics (Vicon 370)	Passive	Video Camera	W	64	2.5K	Yes	Yes	SA (3)	Yes	No	Kn Axis Align	Yes	Sag, Fr, Tr
Peak Performance (Motus)	Passive	Video Camera	W	64	1K	Yes	Yes	SA	Yes	Yes	No	Yes	Sag, Fr, Tr
Qualisys (ProReflex)	Passive	Video Camera	W MAC	16	1.5K	Yes	Yes	SA	Yes	Yes	Optional	Yes	Sag (7)

W = Windows
WNT = Windows NT
Un = Unix
SG = Silicon Graphics
MAC = Macintosh
SA = Semi-Automatic
A = Automatic
Sag = Sagittal
Fr = Frontal
Tr = Transverse
(1) Yes, for 2 or more units.
(2) 2D ID for each camera.
(3) label 1 frame in 1 trial, remaining trials automatic.
(4) Inv/Ev available with CAST & SAFLOU options not in Anatomical option.
(5) No Inv/Ev if Helen Hayes marker set used.
(6) Required with OrthoTrak, optional with KinTrac.
(7) Frontal & Transverse with optional software.

system made in The Netherlands (a company I was unable to contact), the authors tested systems manufactured by five of the companies discussed in this paper (Ariel Dynamics, Inc., Trabuco Canyon, CA; Bioengineering Technology Systems [BTS], Milano, Italy; Motion Analysis Corporation, Santa Rosa, CA; Oxford Metrics Ltd., Oxford, UK; and Peak Performance Technologies, Inc., Englewood, CO). Due to their unavailability, the Japanese and Netherlands systems have not been included in this review.

Active marker systems have LED markers that are pulsed sequentially, so the system automatically knows (by virtue of the pulse timing) the identification of each marker. Marker tracking is not a problem, since the system can maintain the identification of markers temporarily lost from view or with crossed trajectories. Merging of markers can not occur with these systems, so the markers can be placed close together (**Figure 5**). These systems have the disadvantage of requiring that more equipment be placed on the user. A battery pack, pulsing circuitry, and the LEDs and cables must be attached to and carried by the user. For long duration tests, heat generated by the LEDs might be a problem.

Both commercially available systems (**Table 7**), CODA mpx30 (Charnwood Dynamics, Leicestershire, UK) and OptoTrak (Northern Digital, Inc., Waterloo, Ontario) have three cameras mounted in a rigid housing called a "Scanner" (CODA) or "Position Sensor" (OptoTrak). This enables them to be precalibrated at the factory, eliminating the need for the user to acquire calibration data (if only one Scanner or Position Sensor is used). Although the LED markers have a wide

viewing angle, more than one unit may be needed in order to obtain adequate marker coverage for most cinical gait tests. Two units are required to collect bilateral data.

Rather than using conventional video cameras, the CODA mpx30 utilitzes specially designed cameras with a sensing array of photodiodes placed behind a shadow mask with a pseudo-random bar code pattern of black lines. When an LED, on the subject, flashes, a shadow of the mask is cast on the sensor array. The position of the shadow is related to the marker position by straight-line geometry (no lens is used). The averaging effect of signal contributions from all the sensing elements improves the resolving power of the system and provides a high signal-to-noise ratio. The field of view at 4 meters from the scanner is 5-m long × 5.6-m high. Each pair of LED markers is powered by a rechargeable button cell and is strobed by a tetherless IR telemetry system (**Figure 5**). The maximum number of markers—28— can be tracked at a 200 Hz sampling rate.

For the OptoTrak, the field of view at 6 meters from the Position Sensor is 2.6-m long × 3.5-m high.

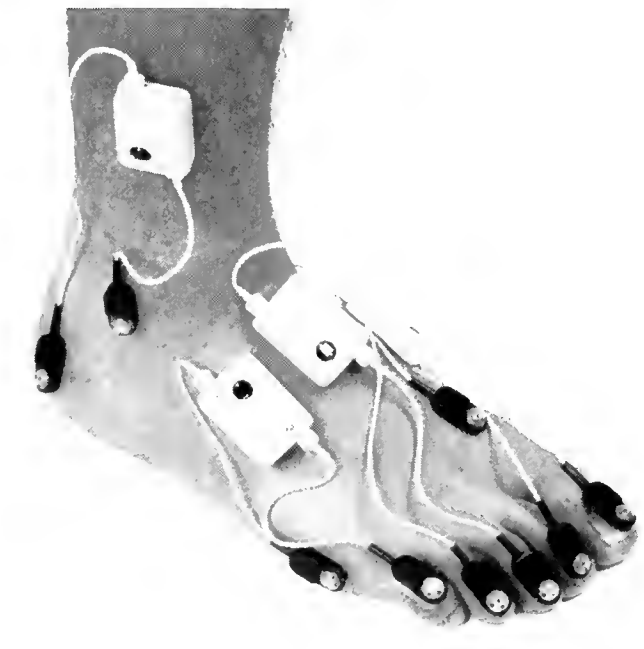

Figure 5.
CODA mpx30 active, light-emitting diode (LED) motion markers placed on the foot of a subject. The sequentially strobed markers enable them to be placed close together without merging in the cameras. Each battery pack provides the power for two markers and houses circuitry to receive an infrared (IR) strobe signal. Photograph is courtesy of Charnwood Dynamics and is used with permission.

An optional tetherless strober is available to eliminate the cable between the wearer and the control unit. The body-worn battery pack (required with the tetherless strober) weighs about one kg. Gait kinematic and kinetic software are not provided. Available software includes a data analysis package, real time rigid body, and application programmer's interface (API) programs. The API software (windows-based) allows clinicians to create their own application programs. The other software is DOS-based.

Passive marker systems have the advantage of using lightweight reflective markers without the need for electrical cables or batteries on the user. IR LEDs around each camera lens send out pulses of IR radiation that are reflected back into the lens from the markers (**Figure 6**). IR filters are used on the camera lenses and system thresholds are set to pick up the bright markers while less bright objects in the background are suppressed. Because of their passive nature, each marker trajectory must be identified with a marker label and tracked throughout the test. When markers are lost from view or their trajectories cross, they can lose their proper identification. Sophisticated tracking software exists that does a good job; however, user intervention is sometimes required. Potential merging of markers in various camera views places limitations on how close together markers may be placed with these systems. The six passive marker systems reported here require the collection of calibration data. Other features vary, but all provide kinematic and kinetic software (**Table 7**).

Laboratory Configuration

Lab configuration for video motion analysis usually ends up being a compromise between optimum camera placement and available space. Manufacturers provide good technical assistance in setting up their respective systems. In general, however, one should keep in mind a few "rules of thumb" to go by:

- Don't try to get by with only two cameras, as there is no way to position them to always have both cameras viewing all markers.
- Make sure the angle between any two cameras is greater than 45°. If two cameras that are separated by a small angle are the only cameras "seeing" a given marker, the determination of the marker's 3D coordinates is less accurate.
- Drape any exterior windows to eliminate outside light from the test area.

- Avoid locating cameras in high trafficked areas where one might be accidentally bumped after the calibration.
- Attempt to keep camera strobe lights from shining directly into cameras across the room.

The last two potential problems can be minimized by mounting the cameras to the walls or ceiling at an elevation of from 6 to 8 feet above the floor; they are then less likely to be bumped and the slight downward viewing angle will minimize strobe light glare from other cameras.

To determine the camera locations for a particular laboratory layout, one would draw a floor plan to scale, drawing a rectangle (around the force plate locations) the size of the desired motion test area. For each camera, a translucent, colored plastic, isosceles triangle should be cut out, with the acute angle equal to the lens-viewing angle. These triangles are laid on the scale drawing of the lab at the approximate desired camera location (**Figure 7**). The triangles are then moved around until each one covers the rectangle representing the motion test area. The camera locations are marked on the drawing and scaled off to obtain the actual camera laboratory coordinates. The system manufacturer should be able to provide the viewing angles for the camera lenses being used. If not, they are easy to measure by moving a marker horizontally in front of each camera and observing its location on the video monitor. One mark is made on the floor where the marker comes into view on one side and another mark where it leaves the other side of the monitor. The angle is measured between the two lines formed by these marks and the camera to obtain the viewing angle.

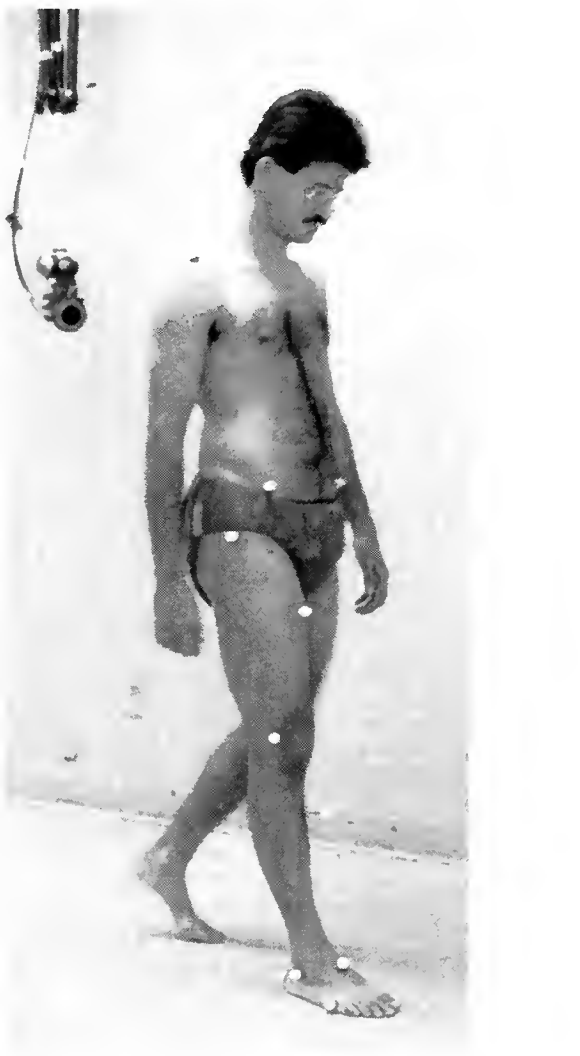

Figure 6.
A subject walking with light weight reflective (passive) motion markers positioned for a unilateral gait test in the Pathokinesiology Lab at Rancho Los Amigos Medical Center. One of six Vicon video cameras (hanging from the ceiling) is visible in the upper left of the photo. Note the ring of IR LEDs around the camera lens. (Photograph used with permission.)

FORCE

Gait is the result of muscle action exerting forces on the skeletal limb segments to produce motion and hence locomotion. It is not possible to measure these internal muscular forces; however, we can learn a lot about pathologic gait and joint loading by measuring external forces.

Force Plates

A force plate measures the ground reaction forces exerted by a person as he or she steps on it during gait. These devices consist of a top plate (mounted level with the surrounding floor) separated from a bottom frame by force transducers near each corner. Any force exerted on the top surface is transmitted through the force transducers. Force plates enable one to measure not only the vertical and shear forces, but also the "center of pressure" during gait. Modern video motion systems have made the determination of joint forces and moments possible through their kinetics software, which requires ground reaction forces.

Two types of force plates are commercially available: piezoelectric and strain gauge. For clinical gait applications, the type probably makes very little difference. Piezoelectric force plates utilize quartz transduc-

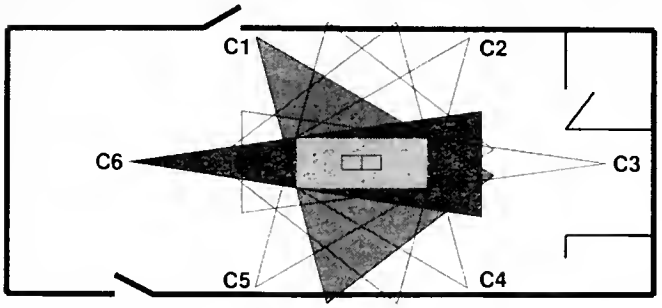

Figure 7.
Gait lab floor plan showing a method of determining camera locations to cover desired test area (light gray). Isosceles triangles represent the viewing angle of cameras labeled C1 through C6. Triangles for C1 and C6 are shown in different shades of gray for clarity. The triangles are moved until they just cover the desired test area. The point of the acute angle then represents the location of the respective camera. Two force plates are shown in the center of the test area.

ers, which generate an electric charge when stressed. They do not require a power supply to excite the transducers; however, special charge amplifiers and low noise coaxial cables are required to convert the charge to a voltage proportional to the applied load. The transducers are calibrated at the factory and no recalibration is necessary. In general, piezoelectric force plates are more sensitive and have a greater force range than strain gauge types. They do have some slow drift, which requires resetting of the charge amplifiers just prior to data acquisition. Strain gauge force plates utilize strain gauges to measure the stress in specially machined aluminum transducer bodies (load cells) when a load is applied. They do not require the special cabling and charge amplifiers of the piezoelectric type; however, they do require excitation of the strain gauge bridge circuit.

Three manufacturers (**Table 8**) produce a large variety of force plates for use in gait analysis. In addition to their own force plates, Kistler also sells the Bertec line.

Force Plate Installation

Considerable planning is required prior to installing the force plates, unless one intends to use unmounted force plates (a technique I do not recommend for gait testing). The top of the force plate(s) must be level with, yet not touch, the surrounding floor. In addition, the mounting structure (e.g., pylon, frames) should be as rigid as possible. Ideally, the force plate(s) should be mounted on a concrete pylon completely separate from the building. In most instances, however, basements or second and subsequent floor installations make this impossible. As a result, I am restricting my remarks to installations on an existing floor. This will require an elevated runway or floor for the gait testing so the force plate(s) can be mounted on the existing floor. An ideal method is to install a "computer access floor," such as the Tate ConCore 1250 access floor[1]. This floor has 610-mm square removable steel floor panels with a cementitous core material, which adds stiffness and minimizes the hollow sound when it is walked upon.

For fixed installations, the force plate mounting frame or frames (available from the manufacturer) should be anchored to the existing concrete floor with threaded anchors and nonshrinkable grout. Different manufacturers have slightly different recommendations and will provide assistance. For multiple force plates, the locations must be carefully planned to provide the widest range of testing. If set up for testing children, the configuration will probably be unsuitable for adults and vice versa. For this reason, many labs have gone to moveable force plate installations.

Movable installations require a large custom-made mounting plate with predrilled mounting holes at discrete locations (anchored similarly to that for fixed installations) or a large flat surface on which air bearing force plate carriers can operate to provide infinite adjustment of force plate locations (17). In this latter technique, Stanhope and Jarrett used an optical bench as the flat mounting surface.

We have adopted this optical bench method in our new gait lab installation (currently under construction). We are using a less expensive 1.07×1.68 m×102-mm thick, model CS-46-4, optical breadboard[2], which has slightly reduced flatness and stiffness specifications as opposed to an optical bench (**Figure 8**). The optical breadboard is being grouted to the concrete floor, with a non-shrinkable grout, to increase the rigidity and stiffness of the installation. Air bearings[3] (Flying Carpet Model "A" Floating Air Platform) are used to levitate the force plates for moving, and magnetic locks[2] (Newport Corp., model 150) lock them into place for testing[4] (**Figure 9**). A simple floor panel cutout scheme has been devised that allows two force plates to be located in multiple configurations (**Figure 10**). Corner

[1] Tate Access Floors, Inc., 7510 Montevideo Rd., Jessup, MD 20794.
[2] Newport Corporation, 1791 Deere Ave., Irvine, CA 92606.
[3] C&H Precision Tools, Inc., 194-20 Morris Ave., Holtsville, NY 11742.
[4] Personal communication with Steven J. Stanhope, PhD, January 27, 1997.

sections, with one dimension equal to the width of the force plates and the other dimension equal to half the length, are removed from four floor panels to accommodate these various configurations.

Advanced Mechanical Technology, Inc. (AMTI) of Watertown, MA, has developed a variation of the infinite adjustment mounting technique, sometimes called the "epoxy lake" method (**Figure 11**). They replace the optical bench or breadboard with a concrete pedestal onto which is flowed a thin layer (approximately 0.635-cm thick) of liquid epoxy. The liquid epoxy is self-leveling and cures to a hard, smooth surface. Air bearing force plate carriers (provided by AMTI) position the force plates to the desired locations (**Figure 12**). Since the epoxy surface is not Ferro-magnetic, magnetic locks can not be used to hold the force plates in place. Due to the weight of each force plate and its carrier (approximately 112.5 kg), it is unlikely that the shear forces developed during walking would cause them to move. Some installations use inflatable bladders resting against the force-plate carriers and the floor substructure to hold the force plates in place. Clinicians contemplating this technique should consult with the force-plate manufacturer to see if that is a viable option for use with the clinicians' force plates.

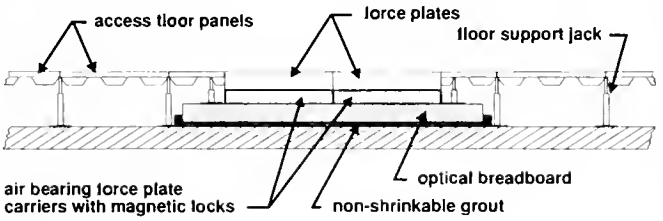

Figure 8.
Movable dual force plate installation on an existing concrete floor (crosshatch), utilizing an optical breadboard as the mounting surface. Force plate carriers with air bearings and magnetic locks position the force plates to the desired locations on the flat smooth surface of the optical breadboard.

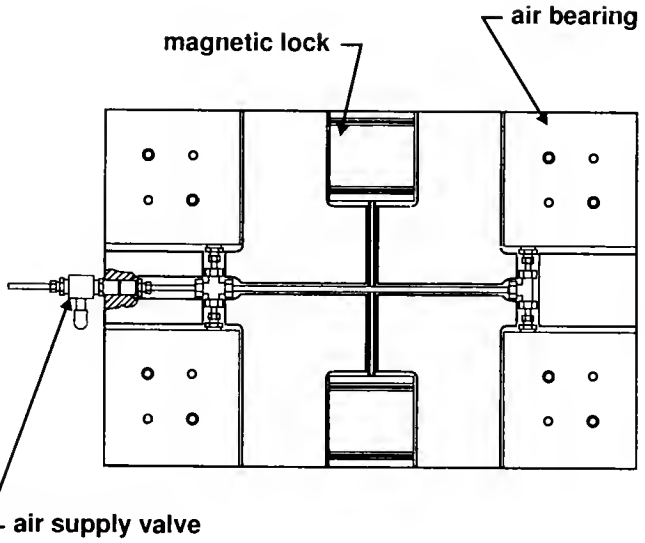

Figure 9.
Bottom surface of a force plate carrier with two magnetic locks and four air bearings. When moving the force plate, the magnetic locks are unlocked and pressurized air is supplied to the air bearings via the air supply valve and tubing. The heavy carrier and force plate can then be easily moved on a thin layer of air.

Force Measuring Sandals

Force measuring sandals record vertical force data from portable transducers attached to the bottom of the feet. They have the advantage of providing multiple strides of data for both feet as the person walks. As in pressure insoles, they do not provide shear forces.

The *Infotronic* (Market-USA, Inc., Severna Park, MD) Computer Dyno Graphy (CDG) system measures the vertical force at eight discrete locations in each of a pair of sandals strapped to the outside of a person's shoes. Factory calibrated capacitive force transducers are fabricated inside 3-mm thick soles. The typical life of a pair of sandals is about 3,000 uses. The wireless system stores the data on small computer memory cards in a data logger (see EMG Acquisition Systems, below). The sandals come in three sizes, small, medium, and large. When walking on a hard surface, the total force is reported to be within 3 to 5 percent of the actual force. There can be a loss of up to 25 percent when the person walks on a carpeted surface. The software provides force time curves, histograms of the force at each transducer, the gaitline (center of force), cyclogram (center of force for both feet), and a listing of the temporal gait parameters.

Force Measuring Walking Aids

Force walking aids are a very valuable tool in determining the amount of load being accepted by the upper limbs during device-assisted gait. Unfortunately, no manufacturer makes force measuring walking aids or load cells designed specifically for insertion in the shafts of canes, crutches, or walkers. However, all three force plate manufacturers sell load cells and both Bertec and AMTI have indicated that they would design and fabricate special load cells if specifications are provided.

Table 8.
Force plate features.

Manufacturer	Type	Sizes	Built in Amplifiers	Software	Special Types
AMTI	Strain Gauge	464 x 508 mm to 610 x 1220 mm	No	Yes	Will custom-make to customer specified dimensions
Bertec	Strain Gauge	464 x 508 mm to 900 x 900 mm	Yes	Yes	All can be used without rigid mounting (1)
Kistler	Piezoelectric	500 x 500 mm to 600 x 900 mm	Yes (1 model)	Yes	Portable system Transparent unit (2) Force Treadmill (3)

(1) As long as shear forces are low enough to prevent slipping of the force plate on the surface.
(2) Allows for photographing through the top of the plate.
(3) Vertical force only, for both feet.

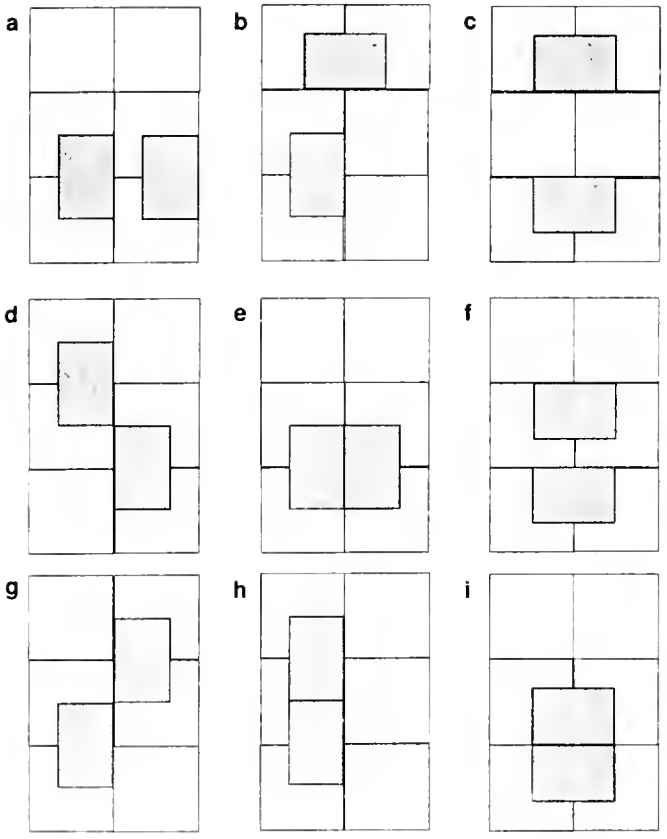

Figure 10.
Floor panel cutout scheme, which allows two force plates to be arranged within a 4-ft wide by 6-ft long area, in 9 configurations (a through i). Four of the 2-ft square floor panels must have one corner removed to accommodate the force plates (shown in gray). One configuration (h) requires two corner notched floor panels and one additional narrow floor panel strip along the left side.

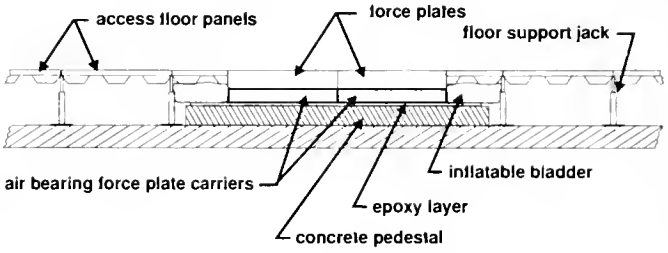

Figure 11.
Movable dual force plate installation on an existing concrete floor with a thin layer of epoxy as the mounting surface. Force plate carriers with air bearings position the force plates to the desired locations on the flat smooth epoxy surface (per AMTI). Although slipping is unlikely because of the weight of the force plates and carriers, inflatable bladders are sometimes used to insure that the force plates will not move.

ELECTROMYOGRAPHY (EMG)

EMG is a valuable tool in clinical gait analysis, as it can give the clinician an accurate representation of what the muscles are doing to contribute to the gait deviations observed and measured by the other instrumentation (i.e., motion, footswitches). Many surgical decisions are made based on the EMG records; therefore, it is extremely important to have instrumentation and techniques that provide high quality EMG signals. Surface electrodes have gained widespread use due to their ease of application and because skin penetration is not required. However, deep muscles can be reliably obtained only with intramuscular wire electrodes, since

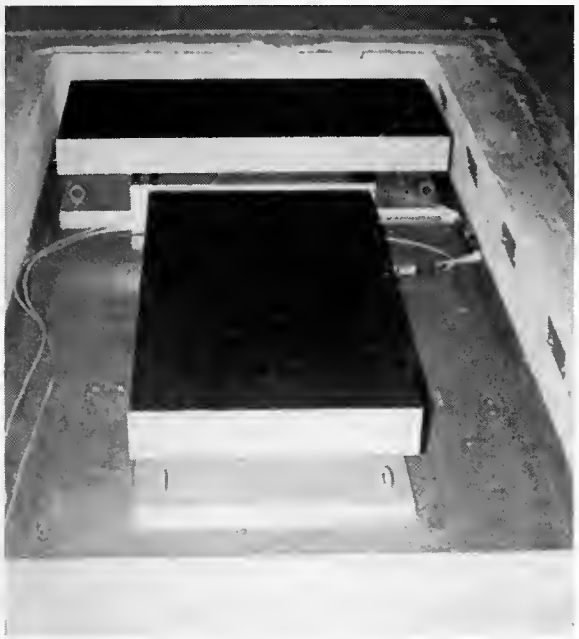

Figure 12.
Two AMTI force plates on their air bearing carriers, resting on a hardened epoxy surface. Floor panels (not shown) are positioned around the force plates for actual gait testing. Photograph is courtesy of Advanced Mechanical Technology, Inc. (AMTI) and is used with permission.

"cross talk" from more superficial muscles will render a surface EMG useless.

Wire EMG Electrodes

EMG Paired Hook Wire Electrodes (Nicolet Biomedical Inc., Madison, WI) are made of insulated nickel alloy wire. The two wires are bent approximately 180° where they exit the tip of a hypodermic needle. The bent end of one wire is 5-mm long and the other 2 mm. Both have 2 mm at the end stripped of insulation. They are available in 25 gauge, 50-mm long and 27 gauge, 30-mm long needles.

Clinicians who choose to fabricate their own intramuscular wire electrodes, can follow the method originally described by Basmajian and Stecko (18) or the modified technique detailed by Kerrigan, et al. (19). We have used the method described by Kerrigan for many years with good results (**Figures 13** and **14**). We use the wire with green insulation, in order to visually tell the difference between stripped and unstripped sections of wire[5] (0.002 Stablohm 800 B Annealed, HPN Insulation, Green). When using the thermal stripper to remove the insulation from the loop of wire extending from the tip of the needle, one must burn all the way up to the needle, as the needle acts as a heat sink, keeping the insulation on the wire inside the needle from charring. This creates a smoother transition between insulated and uninsulated wire at the active end of the electrode. After trimming the two bent ends, the ends must not touch each other under any circumstance.

[5] California Fine Wire Company, 338 South Fourth St., Grover Beach, CA 93433.

Surface EMG Electrodes

Surface electrodes come in two basic types: passive and active.

Passive electrodes are of the "Beckman silver/silver chloride" type and come as individual electrodes, so that a pair can be spaced over the muscle as desired. They are available in various sizes ranging from about 7 mm to 20 mm in diameter (**Table 9**). Conductive electrode gel is required with these electrodes, as well as double-sided tape washers (collars), for attachment to the skin.

Active electrodes have become quite popular, as they provide signal amplification at the electrode site (**Figure 15**). This reduces the electrical "noise," which can be picked up by passive electrode lead wires. A number of electrodes are available, all having high impedance differential amplifier inputs with high common mode rejection ratios. They differ in gain, size, and special features. Two of these electrodes consist of an amplifier package only; therefore, the user must attach separate passive surface or fine wire electrodes to them (**Table 9**). Delsys now manufactures a double differential active surface electrode that is reported to reduce cross talk (20).

EMG Acquisition Systems

EMG data acquisition systems come in two types: cable and wireless. Wireless systems are either radio telemetry or data loggers (**Table 10**). Cable systems eliminate the need for a battery on the wearer (power can be obtained through the cable) and signals are free from any radio frequency (RF) interference or dropout. The disadvantage is the need for a cable connecting the wearer to the instrumentation. Telemetry systems eliminate the cable, but suffer from problems with signal dropout and RF interference. They also require the use of a body-worn battery. Data loggers eliminate the cable and RF problems, but require a body-worn battery and are limited in the amount of data that can be acquired before being downloaded to the computer.

Table 9.
Surface EMG electrode features.

Manufacturer	Type	Electrode Contacts			Ground Reference Electrode	Gain	Bandwidth (Hz)	Can be used with fine wire electrodes?
		Shape	Size	Spacing				
B & L Engineering	Active	Round	11 mm dia.	20 & 30 mm	Separate	330	12 to >1K	Yes
Bortec	Active (1)	NA	NA	User selectable	Separate	500	10 to 1K	Yes
BTS	Active (1)	NA	NA	User selectable	Separate	(4)	(5)	Yes
Delsys	Active (2)	Rectangle	1 x 10 mm	10 mm	Separate	10	DC to 200K	No
In Vivo Metric (IVM)	Passive	Round	7.2 mm dia. to 19 mm dia.	User selectable	Separate	NA	NA	NA
Iomed	Active	Round	10 mm dia.	18 mm	Centered	340	9 to 32K	No
Motion Lab Systems	Active (1)	NA	NA	User selectable	Not required	380	2 to 19K	Yes
Nicolet Biomedical	Passive (3)	Round	20 mm dia.	User selectable	Separate	NA	NA	NA
Sensor Medics	Passive	Round	11 mm dia. & 16 mm dia.	User selectable	Separate	NA	NA	NA

NA = Not Applicable
(1) Amplifier package, only. Used with separate electrodes.
(2) Double differential model available.
(3) Disposable, pre-gelled silver/silver chloride electrode.
(4) Not specified.
(5) See Table 10 for EMG system bandwidth.

EMG Analysis Systems

Much can be learned about a person's gait by a trained clinician viewing the raw gait EMG record; however, computerized analysis systems (**Table 11**) can provide valuable assistance and make the task less tedious and time consuming (21–23). One should keep in mind, however, that computers can only work with the instructions given and the data provided. With patient data, strides can be irregular, and if the software utilizes footswitches to define the gait cycle, problems can occur. For example, a scuff of the foot during swing may appear to the computer analysis software as another stance period. How the software handles these problems is very important. There is no substitute for a trained clinician viewing the raw record to make sure the computer analysis makes sense.

SAFETY (ELECTRICAL ISOLATION)

Electrical safety has always been an important consideration; but with the proliferation of gait labs, it has become an even more critical issue. Any electrical equipment that comes in contact with an individual must be either battery-powered or electrically isolated from the power mains. Electrical isolation is achieved by either transformer or optical isolators. An isolated instrument that is attached to a person should have leakage current of less than 10 micro amps—20 micro amps at the wearer end of a cable connecting the apparatus to that person (24). Not all battery-powered instruments are automatically safe. Consider a battery-powered instrument (on an individual) having data that must be downloaded to a computer. If the interface is

Table 10.
EMG acquisition system features.

Manufacturer	Data Transmission		Number of Channels	Bandwidth/ {sample rate} (Hz)	Filters	
	Wireless	Cable			Highpass (Hz)	Lowpass (Hz)
Bortec		4 mm dia. wire or Fiber Optic cable	8 EMG 2 FSws	10 to 1K	None	None
BTS	FM Telemetry (diversity receiver)	Optional Fiber Optic cable	8 EMG 2 FSws	{5K}	1, 5 & 10	600, 400, 200, none
Konigsberg	PCM FM Telemetry	Optional Fiber Optic cable	8 (any mix of EMG, FSws & other)	{3K}	DC & 2 user defined settings	1K, 500, 250, 125, 62, 32, 16, 8
Market-USA	Data Logger		16 each of EMG, Gonis & FS	{EMG-5K, 200 for Gonis & FS}	None	None
Motion Lab Systems		3 mm dia. wire cable	10 EMG 2 FSws	20 to 2.3K	20 to 170	5, 10, 40, 150, 300, 600, 1.3K, 2.5K
Noraxon	Digitally Encoded FM Telemetry	Optional 10 mm dia. wire cable	8 EMG or 4 EMG & 2 FSws or 4 EMG & 4 Gonis	16 to 500	(1)	500

FSws = Footswitches PCM = Pulse Code Modulated Gonis = Electrogoniometers FS = Force Sandals
(1) A micro chip in the transmitter package implements a specially designed analog, adaptive high pass filter for noise removal. This filter design enables the low cutoff frequency to be very sharp, without ringing.

Table 11.
EMG analysis sytem features.

Manufacturer	Real time Oscilloscope Monitoring?	Normalize EMG?	Display Raw EMG?	Linear Envelopes?	Define GC with FSws?	Onsets & Cessations		Spectral Analysis?	Other
						As %GC?	Compare to Normal?		
B & L Engineering	No	To MMT or max EMG	Yes	Yes	Yes	Yes	Yes (1)	No	(2)
BTS	Yes	To MMT	Yes	Yes	Yes	Yes	Not in US	No	(3)
Market-USA	No	No	Yes	Yes	No	No	No	Yes	(4)
Motion Lab Systems	No	To max EMG	Yes	Yes	Yes	No	No	Yes	(5)
Noraxon	Yes	No	Yes	Yes	Yes	No	No	Optional	

GC = Gait Cycle FSws = Footswitches MMT = Manual Muscle Test
(1) You can incorporate your own data in the normal database.
(2) Report defines whether onset & cessation of EMG was normal, premature, prolonged, or delayed and in which gait phase it occurred.
(3) Database allows tracking of subject groups.
(4) Histogram gives amplitude distribution.
(5) Allows editing of footswitch on and off times.

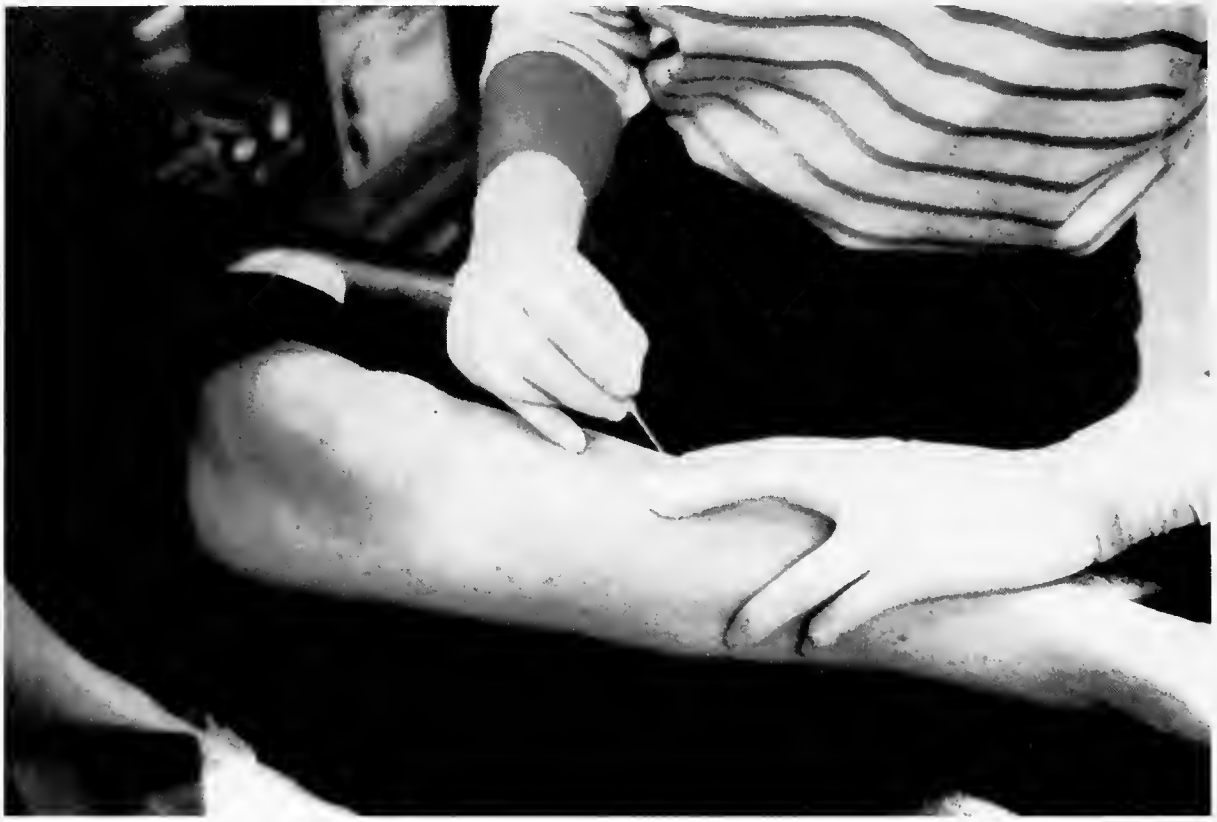

Figure 13.
An intramuscular wire EMG electrode being inserted (with a 25-gauge hypodermic needle) in a muscle of a subject.

not electrically isolated, the package must be removed from that individual before it is connected to the computer. Electrical instruments in the lab, whether they come in contact with the person or not, must be solidly grounded and the ground integrity should be checked on a regular basis. The resistance from the ground prong on the power plug to the chassis should be less than 0.15 ohms. Similarly, the resistance from the ground lead in the power receptacle to a known ground should be less than 0.15 ohms.

All accredited medical institutions have policies and procedures relating to the purchase and safety testing of instruments used in their facilities. Often, Underwriters Laboratory (UL) or Canadian Standards Association (CSA) testing and certification are required. Obviously, small companies manufacturing instruments for a very limited gait analysis market can not absorb the costs of UL or CSA testing. Because of this, many institutions have policies that allow other (less costly) third-party testing. The facility policy manual should be checked to determine what is and is not allowed at that institution.

ACKNOWLEDGMENTS

The author would like to thank Sreesha Rao (for his special help with the motion section) and other staff members of the Pathokinesiology Service, Rancho Los Amigos Medical Center for their helpful suggestions and comments. Bob Manuel and Ron Clark of the Rancho Medical Equipment Repair Shop were very helpful with the safety issues. Thanks, also to the vendors who provided technical information and answered many questions about their respective systems.

Figure 14.
Hypodermic needle being removed from intramuscular wire EMG electrode following insertion in the muscle. Note the loop of wire, which allows the wire to move as the muscle contracts. Used with permission: Craig J. Newsam. Quantification of aquatic therapy water-based methods: Part II: Fine wire electromyography. *The Journal of Aquatic Physical Therapy* 1996; 4(3):13–7.

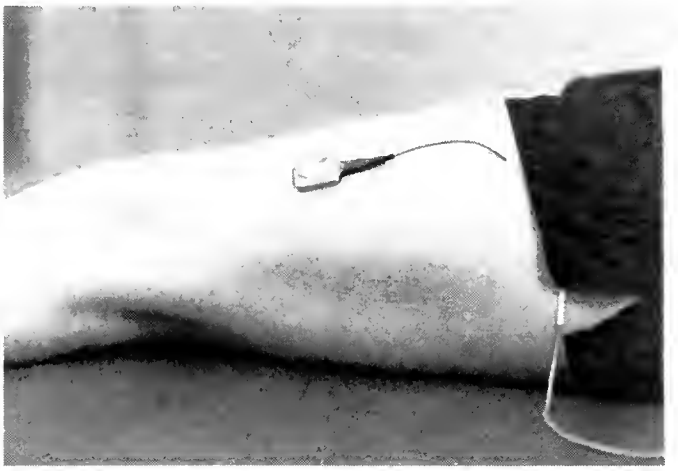

Figure 15.
A Delsys active surface EMG electrode placed over the quadriceps muscle of a subject.

APPENDIX
List of Manufacturers

Advanced Mechanical Technology, Inc. (AMTI)
176 Waltham St.
Watertown, MA 02172 (USA)
TEL: (617)926-6700 (800)422-AMTI
FAX: (617)926-5045
E-mail: lit@amtimail.com
Web: www.amtiweb.com

Ariel Dynamics, Inc.
6 Alicante St.
Trabuco Canyon, CA 92679 (USA)
TEL: (619)874-2547 (714)858-4216
FAX: (619)874-2549 (714)858-5022
E-mail: ariel1@ix.netcom.com
Web: www.arielnet.com

Bertec Corporation
1483 Delashmut Ave.
Columbus, OH 43212 (USA)
TEL: (614)421-2803
FAX: (614)421-2811
E-mail: Bertec@cris.com

Bioengineering Technology & Systems (BTS)
Via Cristoforo Colombo, 1A
20094 Corsico (Milano) ITALY
TEL: +39-2-458751
FAX: +39-2-45867074
E-mail: bts@bts.it
Web: www.bts.it/BTS
US Sales & Support:
TEL: (562)497-1797
FAX: (562)497-1797
E-mail: Fredcei@aol.com

Biometrics Limited (Penny & Giles)
Nine Mile Point Industrial Estate, Unit 25
Cwmfelinfach, Newport
Gwent S. Wales NP1 7HZ (UK)
TEL: +44 (0) 1495 200800
FAX: +44 (0) 1495 200806
E-mail: biometrics_ltd@compuserve.com
Web: www.biometricsltd.com
US Sales & Support:
Penny & Giles Inc.
2716 Ocean Park Blvd. #1005
Santa Monica, CA 90405-5209 (USA)
TEL: (310)393-0014 (310)452-4995
FAX: (310)450-9860
E-mail: nhandler@compuserv.com

B & L Engineering
3002 Dow Ave., Suite 416
Tustin, CA 92780 (USA)
TEL: (714)505-9492
FAX: (714)505-9493
E-mail: sales@bleng.com
Web: www.bleng.com

Bortec Electronics Inc.
7172 Sierra Morena Blvd.
Calgary, Alberta T3H 3G6, Canada
TEL: (403)686-1904
FAX: (403)249-7778
E-mail: bortec@cadvision.com
Web: www.cadvision.com/bortec/bortec.html

Charnwood Dynamics
17 South St., Barrow on Soar
Leicestershire LE12 8LY, England
TEL: +44 1509 620388
FAX: +44 1509 416791
E-mail: support@charndyn.com
Web: www.charndyn.com

CIR Systems, Inc.
790 Bloomfield Ave., Suite 2-10
Clifton, NJ 07012 (USA)
TEL: (973)473-7555
FAX: (973)473-7552
E-mail: sales@gaitrite.com
Web: www.gaitrite.com

Delsys Inc.
P.O. Box 15734
Boston, MA 02215 (USA)
TEL: (617)236-0599
FAX: (617)236-0549
E-mail: delsys@delsys.com
Web: www.delsys.com/~delsys

EQ, Inc.
600 Galahad Rd.
Plymouth Meeting, PA 19462 (USA)
TEL: (215)997-1765
FAX: (215)997-1282
E-mail: jimwalsh@fast.net

IOMED, Inc.
1290 West 2320 South
Salt Lake City, UT 84119 (USA)
TEL: (800)621-3347 (801)975-1191
FAX: (801)975-7366

E-mail: lduffin@iomed.com
Web: www.iomed.com

In Vivo Metric (IVM)
P.O. Box 249
Healdsburg, CA 95448 (USA)
TEL: (707)433-4819
FAX: (707)433-2407

Kistler Instrument Corp., USA
75 John Glenn Drive
Amherst, NY 14228-2171 (USA)
TEL: (716)691-5100
FAX: (716)691-5226
E-mail: biomech@kistler.com
Web: www.kistler.com

Konigsberg Instruments, Inc.
2000 Foothill Blvd.
Pasadena, CA 91107-3294 (USA)
TEL: (626)449-0016
FAX: (626)449-1086

Market-USA (Infotronic)
523 Benfield Rd.
Severna Park, MD 21146 (USA)
TEL: (410)647-2782
FAX: (410)647-5327
E-mail: marketusa1@aol.com

Motion Analysis Corp.
3617 Westwind Blvd.
Santa Rosa, CA 95403-1067 (USA)
TEL: (707)579-6500 (847)945-1411
FAX: (707)526-0629
E-mail: dan.india@motionanalysis.com
Web: www.motionanalysis.com

Motion Lab Systems, Inc.
4326 Pine Park
Baton Rouge, LA 70809 (USA)
TEL: (504)928-GAIT
FAX: (504)928-0261
E-mail: sales@emgsrus.com
Web: www.emgsrus.com

Musgrave Systems Ltd.
Redwither Tower, Redwither Business Park
Wrexham, LL13 9XT (UK)
TEL: +44 (0)1978-66 44 82
FAX: +44 (0)1978-66 44 83
E-Mail: muslabs@aol.com

Nicolet Biomedical Inc.
5225 Verona Road, Bldg. 2
Madison, WI 53711-4495 (USA)
TEL: (800)356-0007 (608)273-5000
FAX: (608)273-6841
E-mail: biomed@nicolet.com
Web: www.biomed.nicolet.com

Noraxon USA Inc.
13430 N. Scottsdale Rd., Suite 104
Scottsdale, AZ 85254 (USA)
TEL: (602)443-3413
FAX: (602)443-4327
E-mail: noraxon@aol.com

Northern Digital, Inc.
403 Albert St.
Waterloo, Ontario N2L 3V2 Canada
TEL: (800)265-2741 (519)884-5142
FAX: (519)884-5184
E-mail: sales@ndigital.com
Web: www.ndigital.com

Novel GmbH
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FAX: + 49 89 337432
E-mail: novel@novel.de
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Novel Electronics Inc.
964 Grand Ave.
St. Paul, MN 55105 (USA)
TEL: (612)221-0505
FAX: (612)221-0404
E-mail: novelinc@novel.de
Web: www.novel.de

Oxford Metrics Ltd.
14, Minns Estate, West Way
Oxford OX2 0JB (UK)
TEL: +44 (0) 1865 261800
FAX: +44 (0) 1865 240527
E-mail: sales@metrics.co.uk
US Sales & Support:
Vicon Motion Systems
15455 Red Hill Ave., Suite C
Tustin, CA 92680 (USA)
TEL: (714)259-1232
FAX: (714)259-1509
E-mail: sales @vicon.com
Web: www.vicon.com

Peak Performance Technologies, Inc.
7388 S. Revere Pkwy. Suite 603
Englewood, CO 80112 (USA)
TEL: (800)PIK-PEAK (303)799-8686
FAX: (303)799-8690
E-mail: peakinfo@peakperform.com
Web: www.peakperform.com

Qualisys, Inc.
148 Eastern Blvd., Suite 110
Glastonbury, CT 06033 (USA)
TEL: (860)657-3585
FAX: (860)657-3595
E-mail: sales@qualisys.com
Web: www.qualisys.com

Sensor Medics Corp.
22705 Savi Ranch Parkway
Yorba Linda, CA 92887-4645 (USA)
TEL: (714)283-2228
FAX: (714)283-8473
E-mail: elizabeth.conner@sensormedics.com
Web: www.sensormedics.com

Tekscan, Inc.
307 West First St.
South Boston, MA 02127-1342 (USA)
TEL: (800)248-3669 (617)464-4500
FAX: (617)464-4266 1 1

REFERENCES

1. Inman VT, Ralston HJ, Saunders JBdeCM, Feinstein B, Wright EW Jr. Relation of human electromyogram to muscular tension. Electromyogr Clin Neurophysiol 1952;4:187–94.
2. Eberhart HD, Inman VT, Bresler B. The principal elements in human locomotion. In: Klopsteg PE, Wilson PD, editors. Human limbs and their substitutes. New York: McGraw-Hill; 1954. p. 437–71.
3. Close JR, Inman VT, Poor PM, Todd FN. The function of the subtalar joint. Clin Orthop Rel Res 1967;50:159–79.
4. Perry J. The mechanics of walking: a clinical interpretation. Phys Ther 1967;47(9):777–801.
5. Perry J. Clinical gait analyzer. Bull Prosthet Res 1974;Fall:188–92.
6. Perry J, Bontrager E. Development of a gait analyzer for clinical use (Abstract). Trans Orthop Res Soc 1977;2:48.
7. Perry J, Bontrager E, Antonelli D. Footswitch definition of basic gait characteristics. In: Kenedi RM, Paul JP, Hughes J, editors. Disability. London: The Macmillan Press LTD; 1979. p. 131–5.
8. Sutherland DH. Gait disorders in childhood and adolescence. Baltimore: Williams and Wilkins; 1964. p. 1–10.
9. Sutherland DH, Hagy JL. Measurement of gait movements from motion picture film. J Bone Joint Surg 1972;54A:787–97.
10. Nelson AJ. Analysis of movement through utilisation of clinical instrumentation. Physiotherapy 1976;62:123–4.
11. Staff of Pathokinesiology Service, Physical Therapy Department. Normal and pathological gait syllabus. Downey, CA: Professional Staff Assoc of Rancho Los Amigos Hosp; 1981.
12. Bontrager E. Footswitch stride analyzer. Bull Prosthet Res 1981;18(1):284–8.
13. Nelson AJ. Functional ambulation profile. Phys Ther 1974;54:1059–65.
14. Murray MP, Drought AB, Kory RC. Walking patterns of normal men. J Bone Joint Surg 1964;46A:335–60.
15. Murray MP, Kory RC, Sepic SB. Walking patterns of normal women. Arch Phys Med Rehabil 1970;51:637–50.
16. Ehara Y, Fujimoto H, Miyazaki S, Mochimaru M, Tanaka S, Yamamoto S. Comparison of the performance of 3D camera systems II. Gait Posture 1997;5:251–5.
17. Stanhope SJ, Jarrett MO. A position adjustable force plate mounting system (Abstract). IEEE Eng Med Biol Soc 1988;10:655.
18. Basmajian JV, Stecko GA. A new bipolar indwelling electrode for electromyography. J Appl Physiol 1962;17:849.
19. Kerrigan DC, Meister M, Ribaudo TA. A modified technique for preparing disposable fine-wire electrodes. Am J Phys Med Rehabil 1997;76:107–8.
20. De Luca CJ, Merletti R. Surface myoelectric signal cross-talk among muscles of the leg. Electroencephalogr Clin Neurophysiol 1988;69:568–75.
21. Bogey RA, Barnes LA, Perry J. Computer algorithms to characterize individual subject EMG profiles during gait. Arch Phys Med Rehabil 1992;73:835–41.
22. Bogey RA, Barnes LA, Perry J. A computer algorithm for defining the group electromyographic profile from individual gait profiles. Arch Phys Med Rehabil 1993;74(3):286–291.
23. Perry J, Bontrager EL, Bogey RA, Gronley JK, Barnes LA. The Rancho EMG Analyzer: a computerized system for gait analysis. J Biomed Eng 1993;15:487–96.
24. Staff. American national standard safe current limits for electromedical apparatus. AAMI standards and recommended practices. 1985 Reference ed., Arlington, VA: Association for the Advancement of Medical Instrumentation; 1984. p. 219–34.

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SECTION TWO

Chapter One

The Contribution of Dynamic Electromyography to Gait Analysis

by Jacquelin Perry, M.D.

Dr. Perry is Director of the Pathokinesiology Lab at the Rancho Los Amigos Medical Center in Downey, California.

INTRODUCTION

The purpose of dynamic electromyography is to accurately define the muscle action that controls joint motion. While gross function of muscle groups can be inferred from motion and moment calculations, specificity of muscle function requires a more discriminating technique.

The Functional Challenge

Walking relies on selective timing and intensity of appropriate muscles at each joint to provide weight-bearing stability, shock absorption, and progression over the supporting foot during stance and to advance the limb in swing. Energy is conserved by activating only the muscles optimally aligned for each task and by substituting momentum and passive tissue tension for direct muscle activity wherever possible.

Throughout this sequence of functions, the muscles perform in groups, as shown in **Figure 1** (1). While the dominant motions of the lower limb occur in the sagittal plane (i.e., flexion and extension for the demands of progression), there also are significant actions in the other two planes (coronal and transverse) to enhance single limb balance and body rotations. Each muscle has a unique three-dimensional (3D) effect determined by its alignment across the joint or joints it crosses. In addition, most muscles are members of two or more functional groups. This redundancy assures 3D balance and serves to simplify the integration of adjacent joint action. Relative intensity of action of a particular

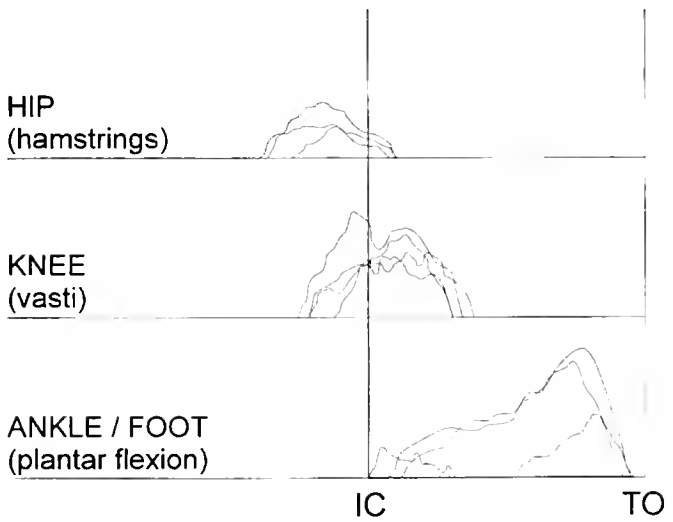


Figure 1.
Normal sequence of synergistic activity of the major extensor muscle groups during stance. Linear display of the EMG amplitudes (vertical scale) of the individual muscles identify their relative intensity and timing. Hamstrings (biceps femoris, semimembranosus, semitendinosus); Vasti (intermedius, lateralis, medialis longus, medialis oblique); Plantar flexors (soleus, gastrocnemius, tibialis posterior [biphasic]). IC (initial contact) indicates onset of stance. Note extensor muscles begin in late swing.

muscle is determined by which of its functions is momentarily dominant. Hence, just understanding normal function requires a detailed study of individual muscle action. Such information alsocan identify the effects of orthoses, muscle training regimens, etc. Dynamic electromyography offers the means of precisely relating muscle action to the specific function.

The Influence of Pathology

The normal, complex walking pattern can be disrupted in many ways. Muscles may be weakened by disuse, pain, or direct injury. Fibrous tissue contracture may limit passive mobility. Orthoses incidentally restrict adjacent motion while purposefully protecting the area of concern. Brain and spinal cord injury may disrupt the primary motor control and feedback pathways. Persons with spastic paralysis, stroke, or head trauma, present the greatest diagnostic challenge as muscle function is disrupted at many levels and the overlay of spasticity often causes the clinical tests to differ significantly from the muscle pattern used during walking. Even lower motor neuron lesions can present unpredictable situations. Individuals preserve their ability to walk by substituting, to the extent their selective control allows. Alternate motions and muscle actions are used to overcome the limitations imposed by pathology. Such substitution capability varies markedly among individuals. Consequently, the person's walking pattern is a mixture of primary functional loss and substitutive actions. The results are mixtures of inadequate, excessive, inappropriately timed, or out-of-phase muscle action. To best design retraining protocols, optimize orthotic assistance, or to plan an appropriate reconstructive surgical procedure, it is essential to know muscle function as it is occurring rather than assumed. This requires dynamic electromyography.

METHODS

Myoelectric Signal Anatomy

Electromyography (EMG) is a system that records the electrical signals activating the muscle fibers. From such information, one can determine the timing and relative intensity during both normal and abnormal function. Under specific circumstances, muscle force also can be calculated.

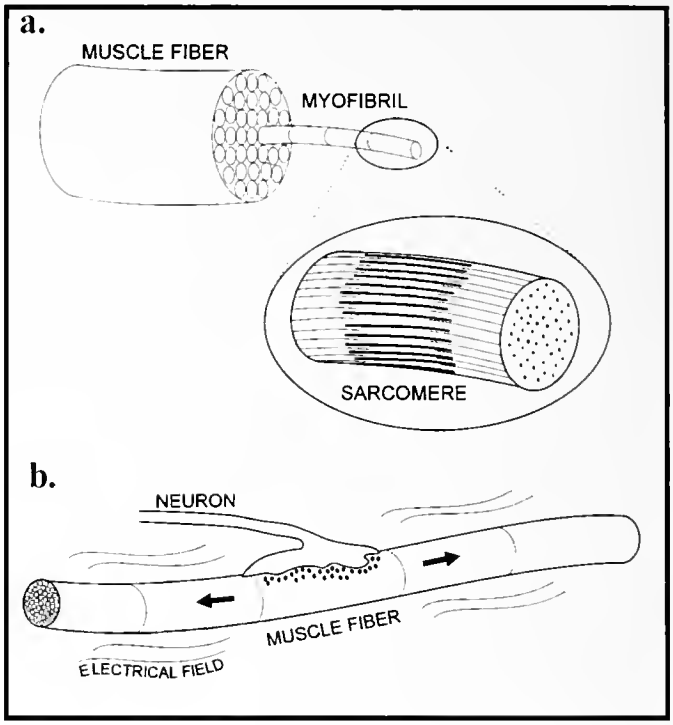


Figure 2.
a) Muscle Fiber Structure: Each muscle fiber is a bundle of myofibrils (chains of contractile units called sarcomeres). Interplay of the thin and thick filament within the sarcomere creates the muscle force. b) Muscle Fiber Myoelectric Signal: An electrical field is created by stimulation from the neuron activating the muscle fiber's chemical receptors (shaded circles), which in turn, send an electrical charge up and down each myofibril to activate the chain of sarcomeres. Adapted from reference (2). Used with permission.

Each muscle fiber consists of multiple long chains (myofibrils) of contractile units (2) called sarcomeres, which create the force of muscle action (**Figure 2a**). As the local neuron chemically activates the muscle fiber at its myoneural junction, an electrical charge is sent up and down each myofibril (**Figure 2b**), stimulating the sarcomeres to contract (3). This event creates an electromagnetic field, which can be used to track muscle activity (4). By volume conduction, the local signal spreads through the tissues making it technically possible to record the signal at the skin surface as well as internally.

Neural control is simplified by having large groups of muscle fibers controlled by a single motor cell body located in the anterior horn of the spinal cord. This

composite of cell body, connecting neuron, and the muscle fiber cluster is called a motor unit. The gastrocnemius, for example, is composed of approximately one million muscle fibers clustered in 600 motor units (5). Animal experimentation has shown that the muscle fibers of each motor unit are widely dispersed through the muscle. Only a few units are needed to create a weak effort throughout the whole muscle. In the multipennate soleus, for example, one motor unit is spread across 60 percent of the muscle's volume, as shown in **Figure 3** (6). Theoretically, just two motor units would be sufficient to traverse the whole muscle. In contrast, a motor unit in the unipennate tibialis anterior covers only 16 percent of the volume (7). Now 6 motor units would be needed. The practical interpretation of this anatomical fact is that during walking and other physiological functions, muscle action can be recorded regardless of the location of the electrode over or within the muscle.

Interspersion of tendonous tissues, however, reduces the concentration of muscle fibers; thus, the middle of the muscle belly is the site where the largest signals are obtained. To be even more precise, maximum signal occurs at the muscle's motor point (8). Us ing the gastrocnemius as an example, 6 motor units would represent only 0.1 percent muscular effort, while a clinical strength grade of 2 (poor), which represents a muscle too weak to accept even the resistance of gravity, averages 5 percent. Theoretically, this represents 30 motor units, a minimum contraction situation. As more motor units are activated, the intensity of the muscular response increases and the EMG signal becomes larger. Clinically, this is reflected as a greater functional force.

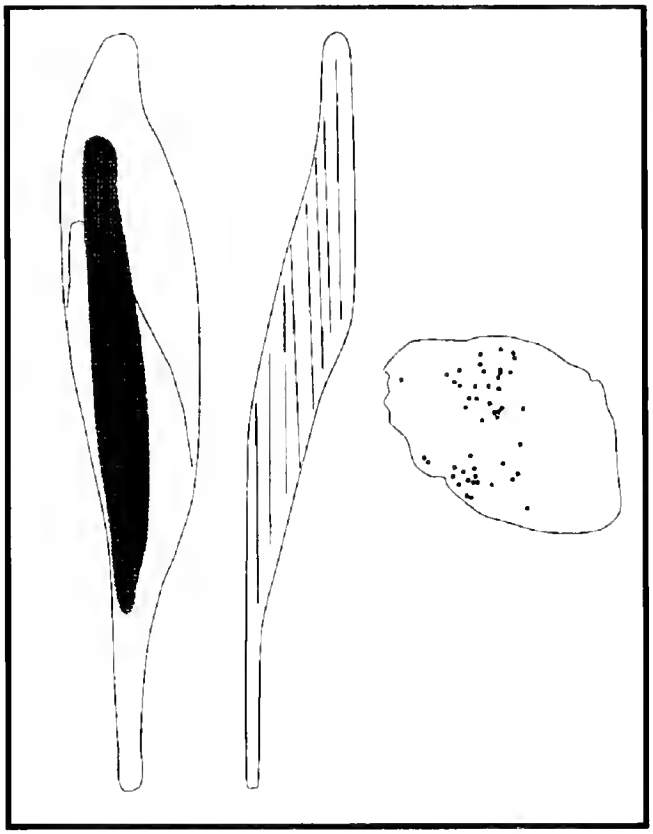


Figure 3.
Motor Unit Territory: The vertical shaded areas in the anterior and lateral projections show the distribution of the muscle fibers of one motor unit within the rat soleus. The cross section identifies the individual muscle fiber distribution (dots) for that motor unit. Adapted from reference (6). Used with permission.

Myoelectric Signal Qualities

The signal recorded during functional EMG is described as random because it does not have a consistent waveform. Instead, the individual spikes vary in amplitude and duration without an identifiable sequence. This inconsistency reflects the fact that every muscular effort is a composite of multiple motor units, each activating multiple muscle fibers. In addition, each fiber's response to stimulation is a brief twitch and, thus, repeated stimulation is required to generate a useful force. Hence, the EMG signal of muscle action is a train of randomly shaped action potentials. In addition, the raw recorded electronic signal is contaminated by noise (i.e., unwanted signals arising from tissue motion and the environment, such as lights, neighboring motors, and so forth). The unwanted electronic noise is excluded by filtering and the use of differential amplifiers, which reject common mode signals.

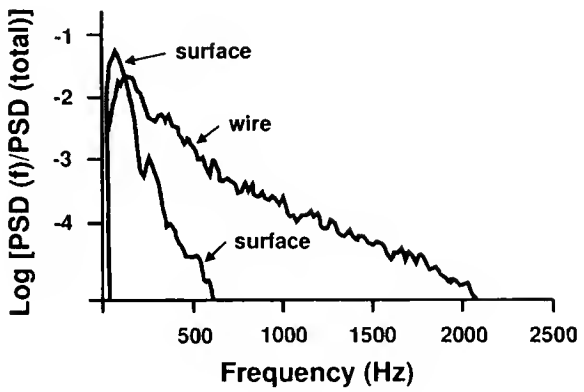
Waveforms are classified by their content of different sine wave frequencies—Fourier analysis (4). In simplistic terms, sharply peaked waves have a high frequency while broad waves have a low frequency. The complex nature of myoelectric signals includes a very

broad spectrum of frequencies, with the range from 10 Hz to 1,000 Hz being considered significant to identify muscle function related to joint motion (**Figures 4a** and **4b**). Tissue displacement accompanying a muscle contraction can generate 10-Hz signals and floor impact during walking gives rise to signals of 25–30 Hz. Hence, 40 Hz has become a customary lower value for gait EMG. In addition, a notch filter is used to exclude the common 60-Hz signals from electrical equipment. Signals above 1,000 Hz do exist but they represent less than 1 percent of the signal power and add nothing to our knowledge of muscle function, so instrumentation with this capability is unnecessary. Hence, a bandwidth of 40–1,000 is appropriate.

Muscle Specificity: Surface versus Wire Electrodes

For functional EMG, the sensing electrode may be either surface contacts (**Figure 5a**) or penetrating wires (**Figure 5b**). The criteria for selecting an appropriate electrode include the purpose of the EMG recording, muscle anatomy, signal dispersion through the tissues, and tolerance of skin penetration with a fine needle.

a.



b.

	FREQUENCY (Hz)		
CUMULATIVE POWER (%)	10	50	90
MUSCLE			
Vastus Lateralis (wire)	130	360	580
Vastus Lateralis (surface)	20	50	100
Sartorius (wire)	110	240	650
Sartorius (surface)	20	40	80

Figure 4.
The typical normalized power spectral density (PSD) for wire and surface electrodes: a) Percent total signal power per frequency interval. Determined by direct Fourier transform of data digitized at 5000 samples/sec, total spectrum 5-2500 Hz with a 5 Hz resolution; b) Thresholds of power spectrum distribution. Frequency below which 10, 50, 90% of the power spectrum occurs. From reference (9). Used with permission.

Surface Electrodes

These EMG sensors have the advantage of convenience and comfort. An active electrode system merely needs to be taped over the center of the target muscle. Passive disc electrodes require a gel and skin cleansing to improve signal transmission. Of the 28 major muscles controlling each lower limb that can be delineated by EMG, the majority are superficial. The dominant period of activity of these subcutaneous muscles can be readily identified by surface electrodes.

The major disadvantages to surface electrodes are cross talk and low signal reception. Their adverse effects complicate the definition of muscle timing and the relative intensity of the activity.

Cross Talk

During periods of low muscle activity, there is the possibility that the EMG record may include signals from musculature other than the muscle of interest. Surface electrodes sense all the signals that reach its reception area. Volume conduction allows wide dispersion of the myoelectric signals through the tissues (10). The thin films of fascia between adjacent muscles present no significant barrier to the myoelectric signals from nearby muscles. Also, muscles function in groups rather than in isolation. As a result, the recording from a surface electrode, by picking up the signals of a synergist may indicate activity in the designated muscle when actually it is quiet.

Several investigators have documented the presence of cross talk by comparing the output of wire and surface electrodes. Perry et al. (11) confirmed group muscle action by demonstrating simultaneous activity of the soleus, gastrocnemius, and tibialis posterior during traditional manual strength tests purported to isolate the targeted muscle. Peak muscular effort, however, corresponded to the designated muscle. The finding that the surface electrodes included EMG from the adjacent muscles implied greater activity than was confirmed by the wire data. De Luca and Merletti (12) studied the signal spread that accompanied electrical stimulation of the tibialis anterior. They found signals in the peroneus brevis and soleus that approximated 17 percent of the maximum tibialis anterior EMG. Koh and Grabner (13), using both stimulation and voluntary quadriceps activa-

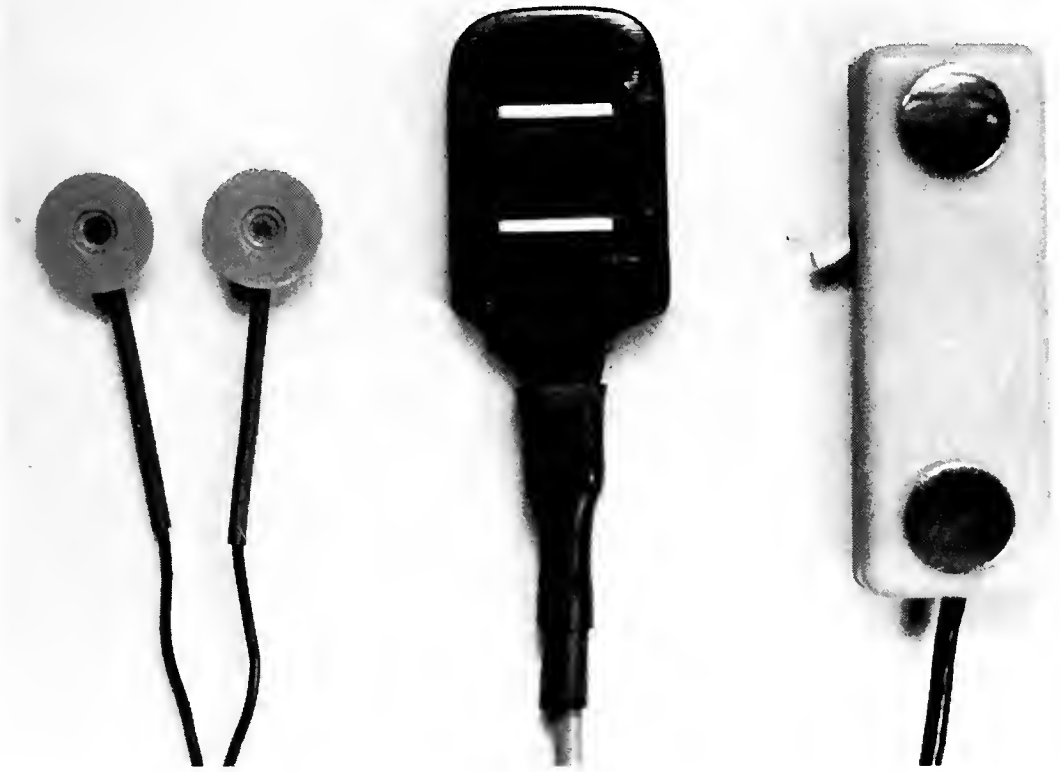


Figure 5a.
Electrodes for Dynamic EMG. Surface: (left) A passive electrode pair containing 2-mm diameter silver silver-chloride disc centers. (Center and right) Examples of active electrodes with signal preamplification circuitry imbedded in the electrode housing. The elements of the center electrodes are 1-cm by 0.1-cm bars spaced 1 cm apart. The right electrode elements are 1-cm discs with an interelectrode spacing of 3.5 cm.

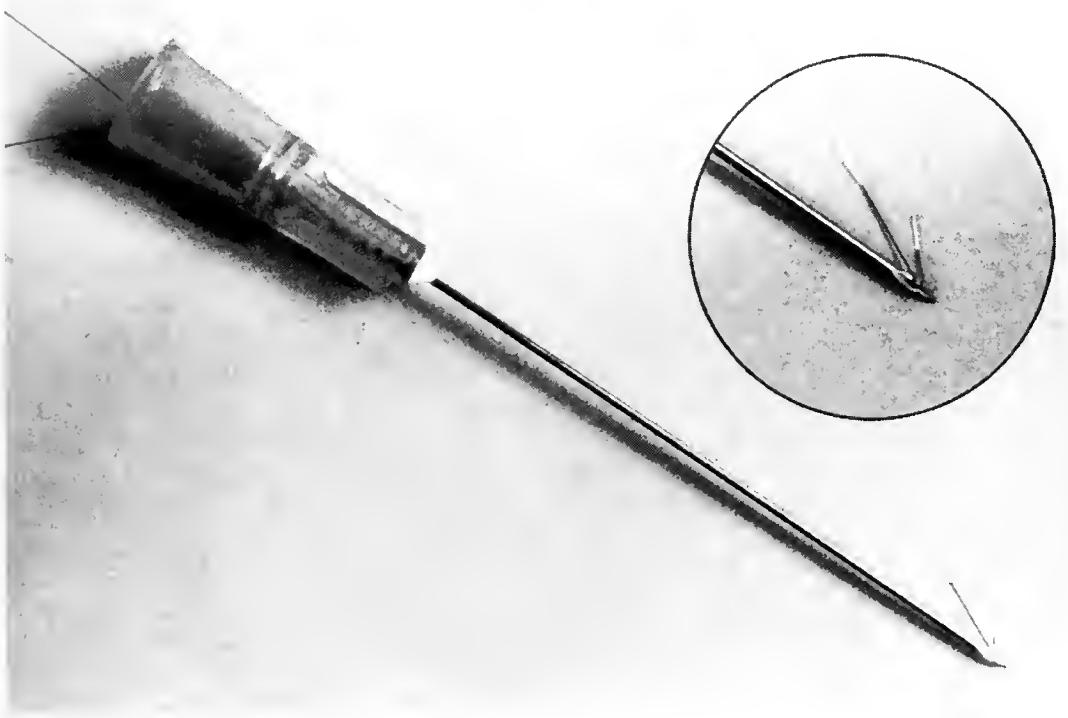


Figure 5b.
Intramuscular wire electrodes are a pair of 50-micron, nylon insulated nickel-chromium alloy wires[1] with the distal 2-mm bare tips, placed in a 3.81 cm 25- or 30-gauge needle for intramuscular insertion. **Inset:** Note, to allow single needle insertion, the external barbs must differ in length to avoid contact between the bared tips.

[1]California Fine-Wire Company, Grover Beach, CA 93433.

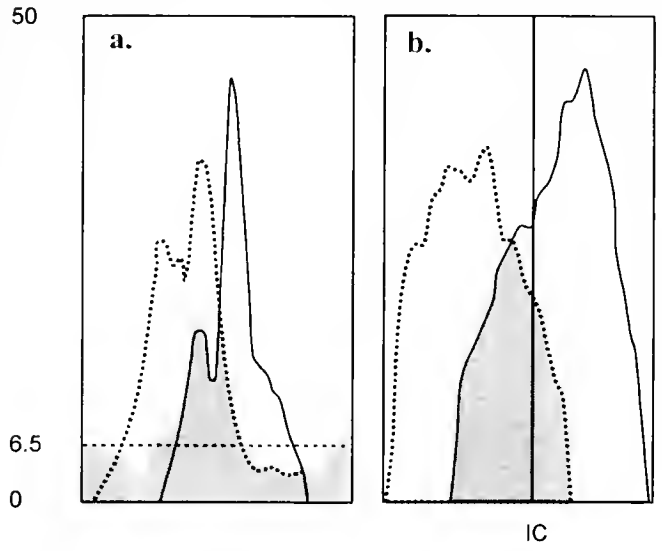


Figure 6.
Cross talk: a) Surface electrode recording of antagonistic flexor (.....) and extensor (___) muscles during walking to identify cocontraction. Shaded area identifies occurrence of simultaneous EMG by both electrodes. Adapted from reference (14). b) Wire electrode recording of similar hamstrings (.....) and quadriceps (____) action. Note the continuous baseline of EMG in the surface recording that is not present in the wire electrode data. This is a display of signal cross talk from adjacent muscles. The taller shaded areas in both recordings represent true cocontraction of antagonist muscles. Adapted from reference (1). Used with permission.

tion to study cross talk, found EMG signals in the medial and lateral hamstrings averaging 11 percent and 17 percent of a maximum effort, respectively. They attributed the difference in these means to the greater distance of the medial electrode from the quadriceps muscle mass. Hence, a significant level of cross talk from adjacent and even moderately remote muscles has been confirmed for both the thigh and lower leg. This complicates the determination of onset and cessation times of muscles' action; thereby confusing the precise identification of muscle phasing, which is a common clinical objective.

At present, there is no established method for circumventing these data complications. Research studies have demonstrated that double differentiation can reduce the cross talk to half or less (12,14). The necessary instrumentation, however, is just becoming available for use in the multiple muscle studies conducted clinically. Faced with this limitation, a possible pragmatic approach might be to eliminate the low intensity signals representing 17 percent of maximum or from 7 to 10 percent of a typical submaximal peak intensity. This could clarify some of the phasing interpretations.

Cocontraction

The interpretation of simultaneous EMG in an agonist and antagonist may be confounded by the presence of cross talk. As Koh and Grabiner concluded, low-to-moderate signals recorded with surface electrodes may be a cross talk artifact rather than cocontraction (13,14). This was demonstrated in a recent study of cocontraction of antagonists in children (15). The authors showed continuous EMG throughout the gait cycle. Superimposed on an average 6.5 percent maximum intensity baseline were regularly interspersed peaks of 20 percent maximum (**Figure 6**). Wire electrode recording from the literature shows that the hamstrings and quadriceps normally overlap in their functions only during limb loading (1); hence, true cocontraction was phasic not continuous.

Wire Electrodes

Intramuscular placement of the EMG sensors circumvents the specificity limitations of surface electrodes. By having the electrode located within the target muscle, a much stronger signal is obtained and its frequency content is higher (**Figure 7**). Both qualities serve to virtually eliminate the problem of cross talk. While myoelectric signals from neighboring muscles may still spread through the tissues, their intensity is insignificant due to their distance from the electrodes.

A second advantage of wire electrodes is the opportunity to use the same signal gain for all muscles. A gain of 1,000 with wire electrodes provides a strong signal for all muscles. This allows the clinician to visually estimate the relative intensity of one muscle's action compared with the others. In contrast, the low reception of surface electrodes (**Figure 7**) commonly requires increasing the gain many fold to obtain a readable signal and the cross talk signals would be similarly magnified. Variability in soft tissue resistance also often necessitates adjusting the gain for individual muscles in order to obtain a readable signal. Thus, wire electrodes allow precise differentiation in the activity of adjacent muscles, making this technique preferable for surgical decisions.

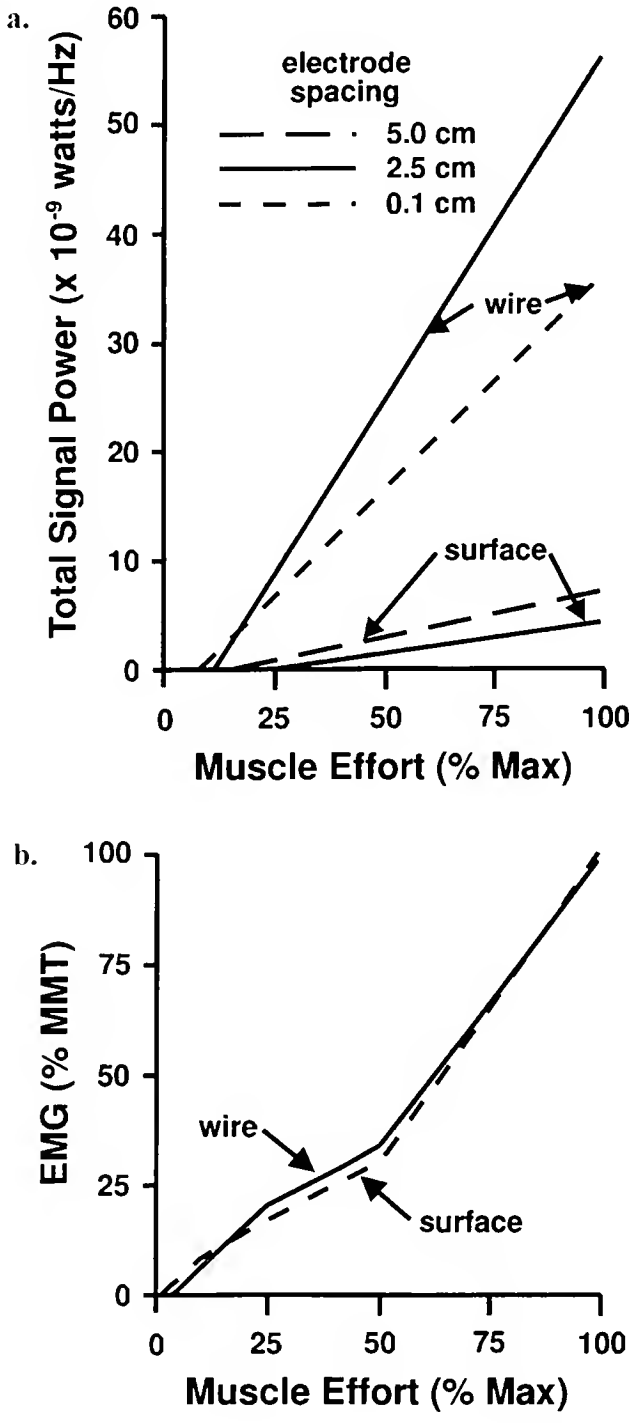


Figure 7.
a) Total signal power of wire and surface with different spacing between the paired electrodes. Wire 2.5-cm spacing inserted with separate needles. Wire 0.1-cm spacing represents single needle insertion. Surface 2.5- and 5.0-cm spacing indicates distance between the centers of two 1-cm diameter discs. b) The Effect of Normalizing: For each electrode (wire and surface), the EMG recorded at each effort level (%Max) was expressed as a percent of the EMG obtained during the isometric maximum muscle test (MMT). From reference (16). Used with permission.

The disadvantage of wire electrodes is the need for skin penetration as the wire electrodes are inserted into the muscle with a fine needle (gauge 25-30). Unless the subject has a bleeding tendency (which would contraindicate wires), the only penalty is momentary discomfort. This is minimized by tensing the skin, knowing the desired location and making a rapid insertion. Children as young as 4 years of age can be successfully tested with wire electrodes. Basmajian and Stecko's technique of inserting both wires with a single needle has simplified electrode location (17). A critical factor, however, is electrode fabrication. The end of the barbs must be of different lengths so that their bared tips will not contact each other and short-out the signal (**Figure 5b,** inset).

For both electrode systems, it is essential that the location relative to the target muscle be accurately determined. Following electrode application, activity of the target muscle is determined by palpable contraction and/or tension of its tendon during a low effort muscle test. Wire electrodes also allow precise localization by light electrical stimulation through the electrodes. Electrodes must be moved until the desired muscle action coincides with the EMG.

EMG Signal Timing

As each muscle provides a specific function, the basic information to be gained by dynamic electromyography is phasing within the gait cycle. The fundamental question is the time of onset and cessation of each muscle's activity relative to the limb motion. A second common concern is the time of peak effort. To make these determinations, some type of event marker must be included with the electromyographic recording to permit phasing. A similarly timed record of limb function is also needed. By itself, the EMG trace is a meaningless sequence of action potentials.

Event Marker

There are basically two methods of identifying the onset of the gait cycle, the use of a footswitch system or a synchronizing indicator on the visible video, motion, or force recordings. Either approach allows one to designate timing as percentage points within the gait cycle. It is customary to begin with initial floor contact as 0 percent and end with next initial contact as 100 percent. The functional significance is made clearer when the gait cycle is further divided into the functional subphases.

Footswitches offer the most versatile approach. While some normal gait studies use just a heel switch, this is seldom adequate as there is no indicator separating stance and swing. For pathological gait, a minimum of four switches on each foot is needed to accommodate the various modes of floor contact (18). The critical sites are heel, medial and lateral forefoot, and great toe. With this system, the basic phases of gait can be determined. The initial double support period identifies initial contact and the loading response phase. Lifting the other foot (contralateral toe-off) identifies single stance. Mid and terminal stance are distinguished as each being half of single stance. One can also relate the EMG pattern to the duration each foot segment is in contact with the floor. Pathology can alter the heel contact pattern in many ways with heel contact being absent, curtailed, or prolonged. While toe-off is the absolute endpoint of stance, a pathological toe drag may obscure the onset of swing. This not uncommon situation, contradicts using "toe-off" as the start of a gait cycle, which some investigators propose (19).

Timing Interpretation

The accuracy of defining the period of significant muscle function by electromyography varies with the technique used. A gross estimate can be made from the raw EMG tracing. This immediately introduces the question of the minimum significant signal (i.e., how small a signal has functional meaning). Most muscle action begins with small spikes representing preparatory activation of a few fibers prior to an EMG record, which progressively shows greater density and amplitude as the effort builds up to the dominant intensity. At the end of the action, there is a corresponding decrement. The slower the action, the more prominent are these small onset and termination packets. They are absent with ballistic movements. In addition, there may be scattered small spikes between the dominant EMG patterns. The inconsistency of these small spikes and amplitudes too small to represent more than trace function imply that they are inconsequential.

With experience, one can learn to subjectively filter out these small spikes by eye. Kaufman found "good agreement" among experienced therapists if they averaged 10 cycles[2]. Di Fabio(20) found that computer designation with established criteria produced consistent reproducibility of onset times, whereas visual

[2] Personal communication, 1994.

analysis by three experienced therapists showed a 51 percent intra-examiner variability and only a 23 percent consistency among examiners. The Rancho computer criteria exclude spikes, which represent less than 5 percent of the muscle's manual muscle test value, and signal packets, which last less than 5 percent of the gait cycle (21). The purpose is to define meaningful muscle function. A second variable is natural inconsistency in timing between strides. The onset and cessation times from three gait cycles has proved to be representative of average function.

Abnormal Timing

Functionally significant deviations from normal timing may occur independently at either the onset or cessation of the EMG record, or both end points may be abnormal. These deviations have been classed as premature, delayed, curtailed, prolonged, continuous, and out-of-phase activity.

Delayed and curtailed EMG indicate inadequate muscle action. For example, curtailed tibialis anterior EMG shows function is limited to just the primitive flexor pattern during initial swing, while the lack of activity in the loading phase of stance identifies that it cannot accompany limb extension (**Figure 8**). Delayed onset of a muscle's EMG is an indication that activation is stimulated by a stretch stimulus rather than central gait control. For example, delay of gastrocnemius action until late terminal stance implies that the dorsiflexion torque was initially controlled by passive stretch of a contracture (**Figure 9**).

Premature, prolonged, or continuous timing are signs of excessive muscle activity. The usual effect is to oppose or partially inhibit normal motion. Premature onset of soleus EMG in swing is a common finding in persons who are spastic (**Figure 8**). Soleus activation accompanies the onset of the primitive extensor pattern by terminal swing knee extension. The unloaded foot is pulled into equinus, leading to premature floor contact by the forefoot. The functional consequence varies with the vigor of the action. A strong, prematurely active soleus can prevent heel contact with the floor, leading to just forefoot support throughout stance; thereby impairing weight-bearing stability.

Prolonged activity most often is found in the hamstrings and must be differentiated from other causes of persistent knee flexion in stance (**Figure 10**). Also, either or both the semimembranosus and the long head of the biceps femoris may act independently. Differences in their timing need to be clarified.

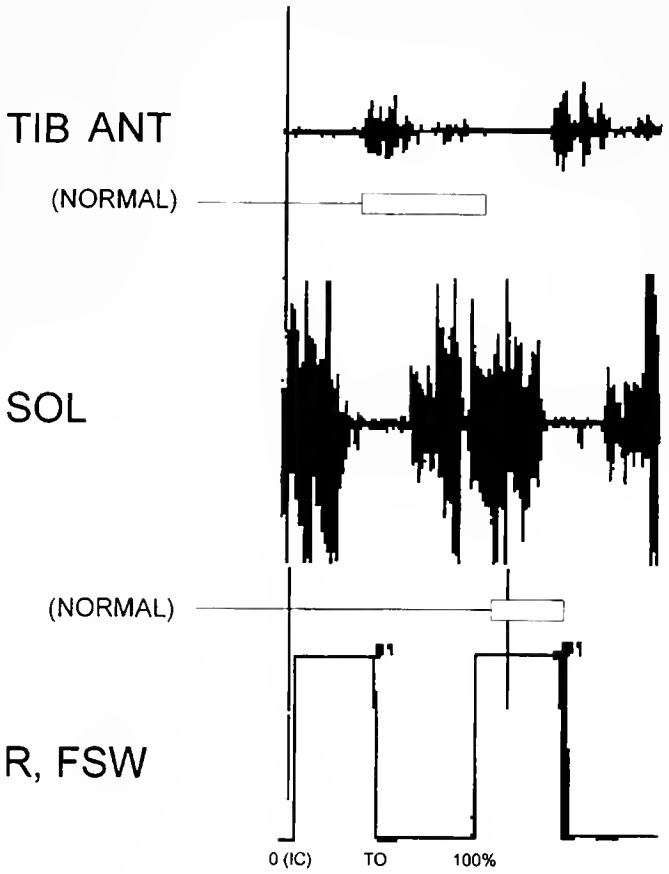


Figure 8.
Curtailed action (Tibialis Anterior): Tibialis anterior onset is appropriate but EMG ceases in mid swing, instead of continuing into the loading response phase of stance. This indicates proper flexor pattern action but inability to contract when the extensor pattern is active. The result is loss of foot support for stance. Premature activity by the soleus: Soleus activity begins in terminal swing rather than after the onset of stance. This implies the presence of a primitive extensor pattern. Footswitches (FSW) pattern identifies forefoot contact only. Diagnosis: cerebral palsy.

Out-of-phase EMG recordings are another form of excessive action. The tibialis posterior may become a swing phase muscle, thereby being the source of excessive foot varus rather than the tibialis anterior. Swing phase quadriceps activity is seen in all types of spastic gait. The effect is obstruction of knee flexion. A major difference among the diagnoses is the source of the obstructive force. Frequently, one or more of the vasti muscles are involved in stroke, head trauma, and spinal cord injury (**Figure 11**), whereas the rectus femoris is the dominant inhibitor of knee flexion in cerebral palsy. This latter situation has led to a

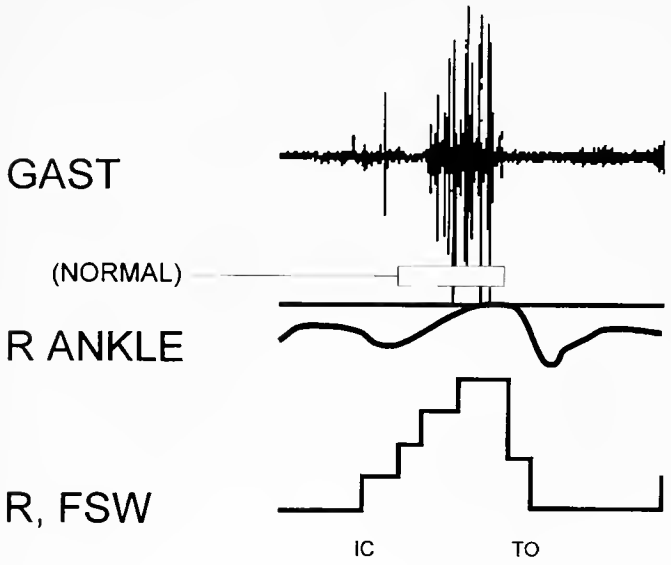


Figure 9.
Delayed onset of gastrocnemius: Ankle goniometer (R Ankle) shows equinus (motion below baseline) at initial contact, which decreases under the stretching force of body weight progression. Gastrocnemius (GAST) EMG onset is delayed until 20% of the gait cycle (normal onset is 5%). This implies contracture tension is the early plantar flexor force prior to stretch, stimulating muscle action. Footswitch (R,FSW) "staircase" identifies stance, baseline is swing. IC= initial contact; TO=toe-off. Diagnosis: post polio.

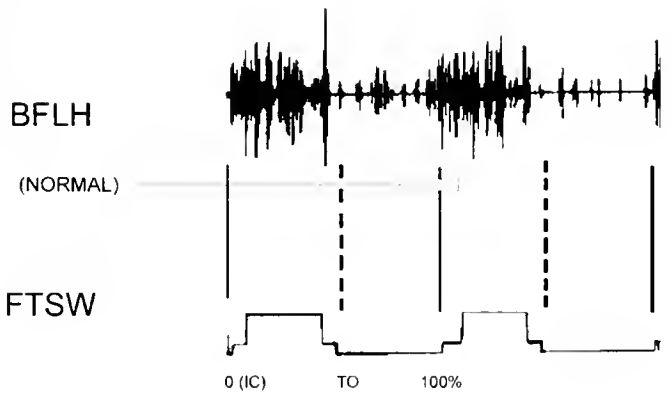


Figure 10.
Prolonged activity of the biceps femoris, long head (BFLH) until late mid stance: The effect was persistent knee flexion in stance beyond the loading response phase that followed initial floor contact (IC). FTSW=footswitch. TO=toe-off. 0 to 100% identifies one gait cycle. Diagnosis: Stroke hemiplegia.

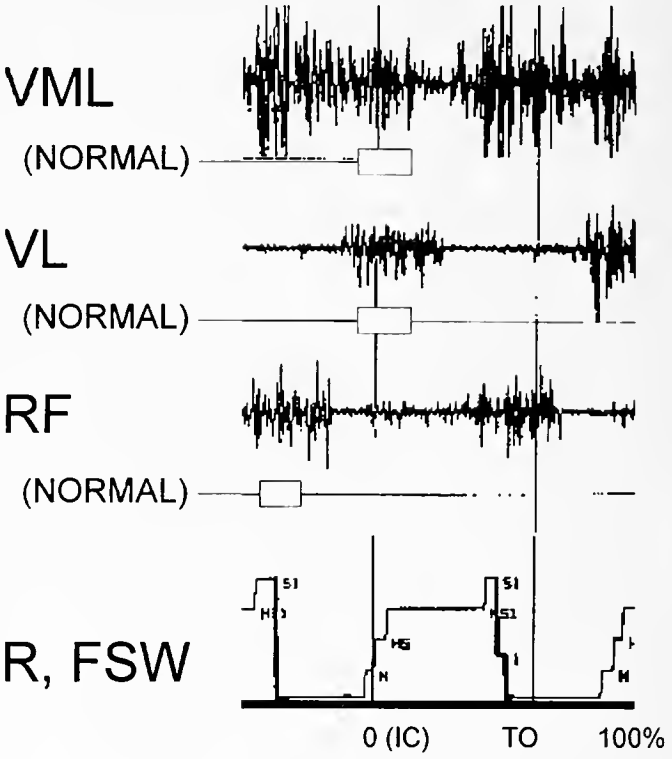


Figure 11.
Out-of-Phase activity of the vastus medialis longus (VML): The continuous EMG identifies swing phase action as well as prolonged activity in stance. Vastus lateralis (VL) displays a nearly normal EMG, identifying spastic muscles have individual sensitivities to stretch and primitive control. Rectus femoris (RF) action is prolonged. Both the VML and RF activity could impede swing phase knee flexion but the more dense EMG indicates that the VML is the dominant inhibitor of knee flexion. R,FSW (right footswitch) designates stance (staircase) and swing gait phases. Subscripts (H,5,1) indicate foot area contacting the ground. Nearly continuous H (heel) contact implies calf muscle weakness. IC=initial contact); TO=toe-off. Diagnosis: Adult traumatic brain injury.

technique of using surface electrodes to identify when the rectus femoris is the cause of limited swing knee flexion (22). For all other diagnoses, intramuscular wires are needed to differentiate rectus femoris action from out-of-phase vastus activity (23).

There are no criteria for the duration of a timing error needed before motion is altered but usually the abnormal timing is quite gross. Superimposed on the timing error is the effect of muscle intensity.

EMG Intensity

Muscles increase their force by the activation of additional muscle fibers or by increasing their firing

rate. Both responses create a more intense electromyogram. Signal amplitude is increased as the simultaneous action potentials add together, while asynchronous potentials form new spikes. Visual inspection reveals an electromyogram with both amplitude and density increased. The level of EMG recorded during gait may or may not be similar to that occurring during the baseline muscle test. Normally, peak gait intensity is approximately a third of the maximum test level. A gait record that exceeds the muscle test is an indication of poor voluntary control. In interpreting the raw clinical record, there are four significant levels of function: absent, inadequate (weak), appropriate (strong), and excessive. Absent gait EMG in a muscle with a notable muscle test value implies that either it is shielded from stretch or being avoided as a detrimental force. Inadequate intensity implies muscle presence but inability to meet the functional demand. Excessive intensity, in the presence of a good muscle test record, is a sign of either obstructive force or muscle overuse and potential fatigue (**Figures 12a** and **12b**). Visual comparisons of relative intensity among muscles are very convenient with wire electrode records, since the same amplification is used for all muscles. With surface electrodes, however, obtaining a readable record generally requires the tester to individually adjust the amplification of each muscle record to overcome the difference in the impedance of the overlying skin and soft tissues. Hence, similar record amplitudes can represent very different muscular effort.

Muscle intensity also can be quantified by either a descriptive scale or computer measurement. A customary descriptive scale uses four intensity levels, with grade four indicating maximum. Small changes, however, are difficult to identify. Today, it is more common to quantify the EMG by computer. This allows fine grading of the muscular effort and accurate discrimination of small differences.

Computer Signal Quantification

Three steps are involved in providing a meaningful numerical value for the muscles' EMG. The raw EMG is rectified, digitized, and normalized (**Figure 13**). Normalization permits the comparison of effort changes among two or more muscles despite the inability to either determine or control the number of muscle fibers that an electrode samples.

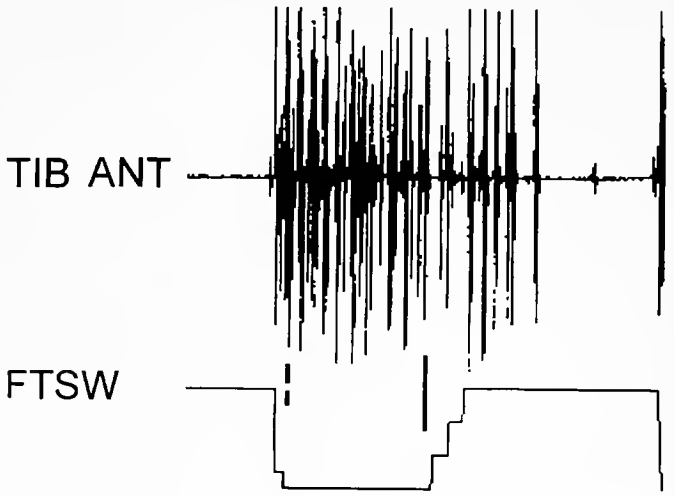


Figure 12a.
The EMG of this tibialis anterior represents a sparse number of enlarged motor units functioning at a higher than normal intensity (quantified as approximately 70% of its muscle test). Excessive action also is evident by the persistence of the same intensity throughout its function phase (swing and early stance). Diagnosis: Post poliomyelitis muscle overuse.

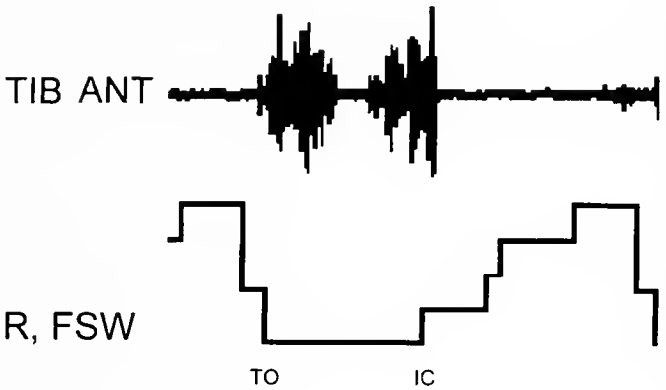


Figure 12b.
Normal tibial anterior is EMG of a young adult: The quick, dense EMG packet provides vigorous dorsiflexion to lift the foot from its plantar flexed position at the onset of swing (TO). Tissue tone is sufficient to support the foot in mid swing (EMG absent). Second burst reactivates dynamic dorsiflexion to support the foot in terminal swing and early stance.

Normalization

To accommodate the need to use uncontrolled EMG samples, all of the EMG values obtained for a given muscle are compared to a normalizing base. Most commonly, this base is the EMG accompanying a

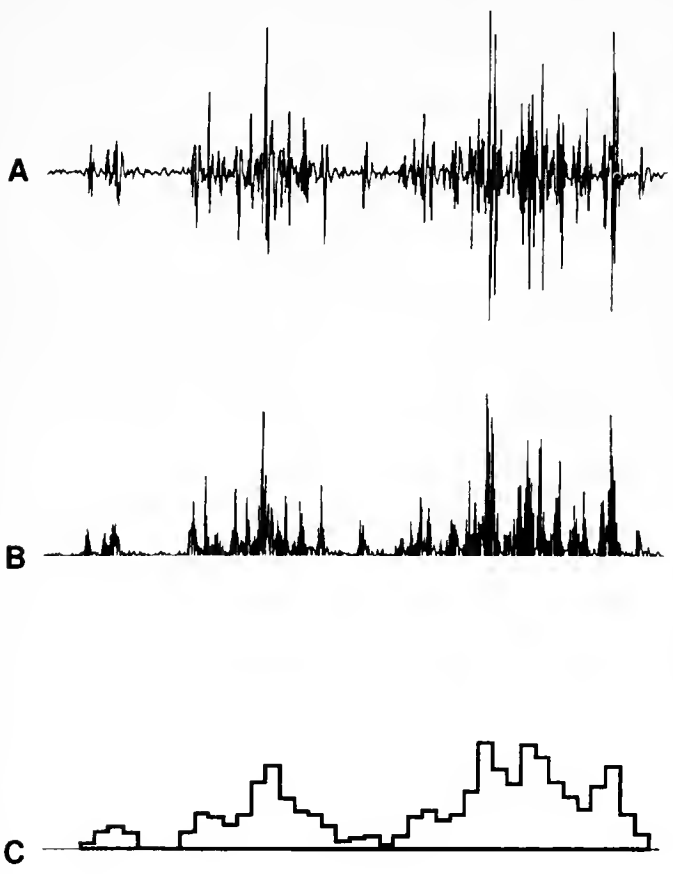


Figure 13.
Computer quantification of EMG record: a) the raw analog data are collected digitally by sampling the signal at 2500 Hz; b) the signals then are rectified by transposing the negative values to positive; and c) the data are normalized and summed over designated intervals (usually 1% of the gait cycle) to generate a linear envelope that expresses the data as percents of the maximum EMG reference.

maximum effort by that muscle. Hence, the individual test values are expressed as a percentage of the base value (i.e., %MVC).

To meet the time constraints of simultaneously testing six or eight muscles in a clinical setting, the manual muscle test maximum is the customary normalizing base (%MMT). The procedure consists of recording the EMG during the maximum effort test, calculating the mean for the one second with the highest EMG, and then relating each functional EMG to that value using a common time interval, generally 0.01 second or 1 percent of the gait cycle.

An alternate approach uses each muscle's peak EMG in the gait cycle as the normalizing base and all other phase values are related to it. This is convenient but it does not allow one to compare relative intensity among muscles, since the peak effort for each is 100 percent. This most often is used in situations where poor patient cooperation makes muscle testing difficult.

Electromechanical Delay

The time between the onset of the myoelectric signal and the initiation of muscle tension is called the electromechanical delay (EMD). This interval is assumed to represent the propagation of the action potential along the muscle, the excitation-contraction coupling process, and stretching of the muscle's series elastic component by the contracting component (8). This delay is significant only if one wants to precisely relate EMG and motion in selected research studies. In general clinical practice, however, the difference in timing is inconsequential. As the following summary identifies, it also involves a very short time period (5,8,24–26).

The differences have been found to relate to three variables: method of muscle activation, mode of recording the signals, and the method of identifying muscle tension. Voluntary effort created the longest delays, and knee extension, which requires moving a larger mass than elbow flexion, was slower. Significantly faster stimulation was attained with a reflex hammer or an electrical current (**Figure 14**). Among the methods of identifying the onset of muscle tension, the slowest was a gross exercise unit, such as a Kin-Com or goniometer (26). A load cell force transducer in intimate contact with the leg registered a quick response (8,27), but the most sensitive motion instrument was an accelerometer. Involved in these differences are both the inertia of the limb and the lag within the mechanical testing system (8). Different effort levels and comparisons of isometric and isotonic action showed only minor differences in the electromechanical delay between onset times, but increasingly higher target forces required proportionally greater total time. The combination of tendon tap stimulation of knee extension measured with a force transducer registered an EMD of 25 ms and electrical stimulation shortened the delay to 20 ms. The shortest EMD (16 ms) was recorded by testing voluntary biceps activation of elbow flexion using an accelerometer for motion sensing and gross magnification of the record for easier reading of the data. It was calculated that the transport time involved only 10 ms (24). Returning to

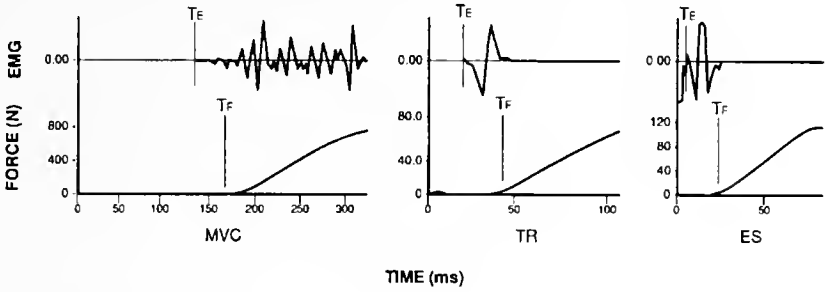


Figure 14.
Electromechanical delay (EMD): Onset timing of EMG and force during three modes of quadriceps (vastus lateralis) activation. Left. Voluntary knee extension (EMD=40 ms); middle. Tendon reflex (EMD=25 ms); right. Electrical stimulation (EMD=19 ms). TE (EMG onset threshold, 0.015 mV), TF (force onset threshold, 3.6N). Note EMD reduced by promptness of muscle activation. Adapted from reference (8). Used with permission.

the question of gait electromyography, the tendon tap response could be likened to eccentric activation during walking. A logical conclusion to draw from these multiple studies is that the average electromechanical delay during gait is no more than 40 ms and perhaps as short as 25 ms or even 10 ms.

EMG Force

Activation of an increasing number of muscle fibers results in a correspondingly greater force. The EMG also increases. The result is a quasi-linear relationship between force and EMG when the muscular effort is isometric but the precise relationship varies with the mode of motor unit recruitment (28). To interpret muscle force from an EMG of different effort levels, however, the data have to be normalized as the ratio (linear slope) between these two factors varies with the muscle studied, electrode placement, and mode of signal recording, and because the number of the motor units sampled and their muscle fiber composition can neither be defined nor controlled (29,30).

Motion markedly distorts the isometric (1) relationship of EMG and force by changing the effectiveness of the muscle fibers, while the EMG continues to identify the relative number of fibers included in the sample. Muscle force (F) is modified by joint position (P), mode of contraction (C), and speed of action (V). The conceptual model may be represented as F=I(V+P+C).

Joint position alters two muscle factors: sarcomere effectiveness and moment arm length. Each muscle fiber is a chain of force units called sarcomeres; within which force production capability is determined by the number of bonds between its myosin and actin filaments. Maximum bonding occurs in the midrange of the sarcomere with force being reduced by either lengthening or shortening of the sarcomere. The length of the sarcomere chain (i.e., muscle fiber) is determined by joint position. Recent *in vivo* studies of wrist extensor sarcomeres have shown that even synergistic muscles (extensor carpi radialis brevis and longus) have optimum sarcomere bonding at different joint positions. Effectiveness of the resulting muscle force in creating motion (moment) is further modified by its functional leverage (moment arm), which also varies with joint position. Optimum sarcomere bonding and moment arm lengths commonly occur at different joint positions, a situation that seems to extend the functional effectiveness of the muscle. For example, quadriceps muscle force is maximum at 60° of flexion (31), but the longest moment arm for the patellar tendon is found at 15° flexion (31).

Muscles have three modes of contraction: isometric (no motion, the dynamic force equaling the passive resistance), eccentric (active resistance to passive lengthening), and concentric (active shortening). The latter two modes are forms of motion. In some muscles, such as the biceps brachii, the eccentric force can exceed isometric capability by 10 to 20 percent. For the quadriceps, isometric and eccentric appear to be similar (32). Eccentric holding by the actin-myosin bonds is

enhanced by titin, a third protein (33). Concentric contraction requires serial re-bonding of the actin and myosin protein filaments as the muscle actively shortens. This is less efficient, resulting in a force approximately 20 percent less than isometric. Hence, for the same EMG signal, the resulting force depends on whether the effort is isometric, eccentric, or concentric; while the EMG representative of muscle fiber involvement remains unchanged (**Figures 15a** and **15b**).

The velocity of motion influences the muscle force of concentric effort but not eccentric activity. As actin – myosin bonding is rate dependent, sarcomere stability is reduced with fast shortening contractions, and muscle force correspondingly decreases. During walking, sarcomere sensitivity to speed relates only to swing phase events. In stance, muscle action is primarily isometric and eccentric; thus, there is a reliable relationship between the normalized EMG and the muscle forces being employed.

EMG Relationship to Moments

During walking, the amount of effort a muscle must exert at any instant in time is determined by the destabilizing influence that falling body weight has on the joint controlled by that muscle. Engineers define this destabilizing rotational force as a moment. The significant factors are the magnitude of the falling body weight force (measured as a ground reaction force) and the perpendicular distance between that force line (vector) and the joint center (moment arm). Stability is preserved by an equal and opposite moment from muscular action. This approach is an accurate representation of normal muscle group function. Antagonistic cocontraction is minimal and there are no other significant destabilizing forces. At the knee, for example, the moment calculation is a good representation of quadriceps effort during weight acceptance as the period of hamstring activity at the onset of stance is brief and of low intensity. At the same time, the mechanics at the foot are contributing to the demand moment. Hence, there are no hidden forces to impose significant deviations in the moment calculations.

A commonly unrecognized problem, however, is the assignment of muscle action to passive events. Contrary to pure mechanics, the human body has an acute feedback system (proprioception), which allows intelligent use of passive mobility. Examples are mid stance hip extension and late stance hip abduction induced by the fall of body weight following the swinging limb (34). In these instances, passive momentum has been used instead of muscle agonists (35). Moment calculations have the added limitation of identifying only group muscle action. Delineation of individual muscle activity necessitates dynamic EMG.

Pathology can impose serious compromises to the prediction of muscle action with moments. In spastic diseases, such as spastic paralysis or stroke, intense cocontraction may exist. Prolonged cocontraction by the hamstring muscles may require greater quadriceps intensity than is indicated by moment calculations. Faulty foot support by either prolonged heel only or forefoot floor contact also can impose unrecognized instability at the knee and hip, leading to additional muscle action not evidenced by the calculated moment. For example, persons with spastic paralysis who have a crouch gait as the result of prolonged hamstring muscle action, preserve balance over their flexed knee by leaning forward. Associated limitations in ankle dorsiflexion impose a toe stance. The resulting posture is accompanied by EMG recordings showing strong quadriceps and antagonistic hamstring activity. Simulation of this posture in nonimpaired subjects confirmed intense cocontraction of agonists and antagonists at both the ankle and knee resulting from limb posture rather than spasticity (36).

SUMMARY

Dynamic electromyography enables the clinician or research investigator to define the timing and intensity of individual muscle function during gait and other functional activities. Moment calculations identify the action of controlling muscle groups during normal function, but may become inaccurate when pathology alters the balance of passive and active forces. Wire electrodes, by their placement within the designated muscle, provide a more precise definition of both timing and intensity of muscle action than do surface electrodes, but require needle penetration of the skin. Surface electrodes have the advantage of convenience.

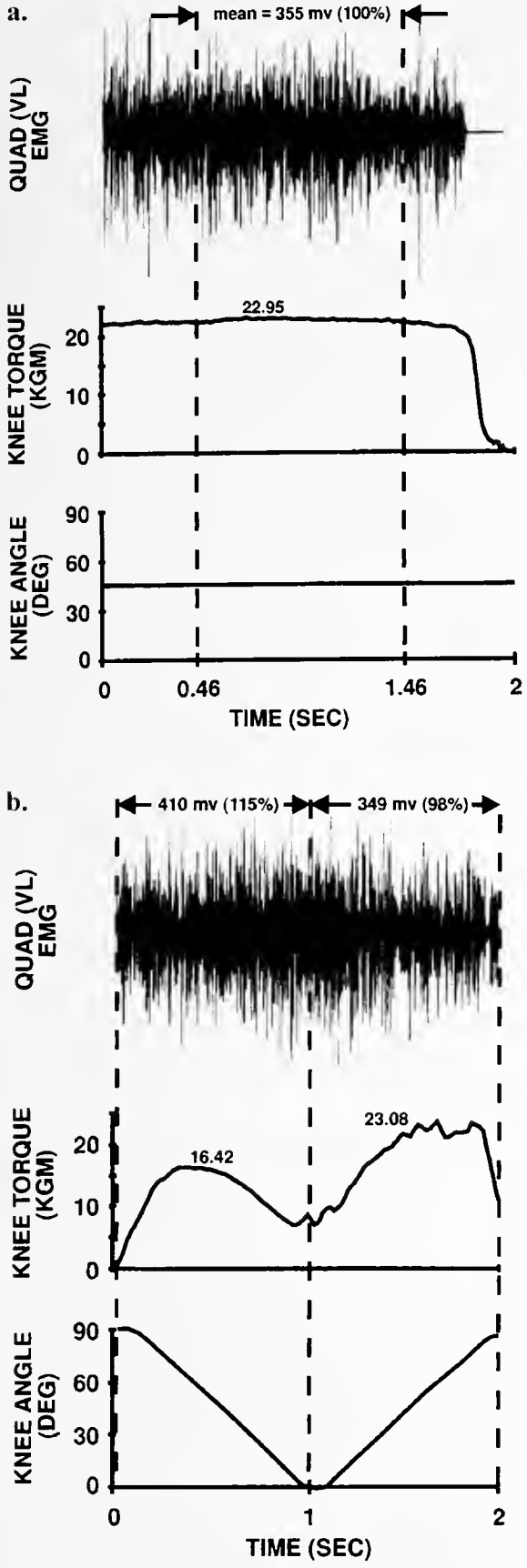


Figure 15.
EMG – Force Relationship per Type of Muscle Contraction: All tests were maximum knee extension and the data were calculated over a 1-second time period. Quadriceps EMG is represented by vastus lateralis (VL). a) Isometric maximum effort at 45° of knee flexion: VL, raw EMG signal and mean intensity (millivolts), also 100%. Torque, analog recording, and peak intensity (KGM, kilogram meters). b) Concentric (left) and Eccentric effort (right). Direction of motion indicated by knee angle pattern. Test arc was between 90° and 0° flexion. Rate was 90° per second. Mean EMG and % isometric were quantified for the 1-second effort in each direction. Torque was calculated as the peak value for 0.1 second. Expression of data as % isometric values showed motion modified concentric force production (EMG 115%, Torque 72%) but not the eccentric effort (EMG 98%, torque 101%).

REFERENCES

1. Perry J. Gait analysis, normal and pathological function. Thorofare, NJ: Charles B. Slack; 1992.
2. Lieber RL. Skeletal muscle structure & function. Baltimore: Williams & Wilkins; 1992. p. 22.
3. Goodgold J, Eberstein A. Electrodiagnosis of neuromuscular disease. Baltimore: Williams & Wilkins; 1972. p. 21–28.
4. Basmajian JV, Deluca CJ. Muscles alive: their functions revealed by electromyography. 5th Ed., Baltimore: Williams & Wilkins; 1985. p. 19–64.
5. Inman VT, Ralston HJ, Saunders JBdM, Feinstein B, Wright EW, Jr. Relation of human electromyogram to muscular tension. Electromyogr Clin Neurophysiol 1952;4:187–94.
6. Burke RE, Levine DN, Salcman M, Tsairis P. Motor units in cat soleus muscle: physiological, histochemical and morphological characteristics. J Physiol(Lond) 1974;238:503–14.
7. Bodine-Fowler S, Garfinkel A, Roy RR. Spatial distribution of muscle fibers within the territory of a motor unit. Muscle Nerve 1990;13:1133–45.
8. Zhou S, Lawson GA, Morrison WE. Electromechanical delay in isometric muscle contractions evoked by voluntary, reflex and electrical stimulation. Eur J Appl Physiol 1995;70:138–45.
9. Freeborn CW, Antonelli D, Perry J. Spectral analysis of EMG signals. Annual Reports of Progress. Downey, CA: Rancho Los Amigos Rehabilitation Engineering Center; 1979. p.35–7.
10. De Luca CJ. The use of surface electromyography in biomechanics. J Appl Biomech 1997;13:135–63.
11. Perry J, Easterday CS, Antonelli DJ. Surface versus intramuscular electrodes for electromyography of superficial and deep muscles. Phys Ther 1981;61:7–15.
12. De Luca CJ, Merletti R. Surface myoelectric signal cross-talk among muscles of the leg. Electroencephalogr Clin Neurophysiol 1988;69:568–75.
13. Koh TJ, Grabiner MD. Cross talk in surface electromyograms of human hamstring muscles. J Orthop Res 1992;10:701–9.
14. Koh TJ, Grabiner MD. Evaluation of methods to minimize cross talk in surface electromyography. J Biomech 1993;26 Suppl 1:151–7.

15. Frost G, Dowling J, Dyson K, Bar-or O. Cocontraction in three age groups of children during treadmill locomotion. J Electromyogr Kinesiol 1997;7:179–86.
16. Perry J, Antonelli D. The dynamics of EMG, force, movement relationships for knee extension. Annual Reports of Progress. Downey, CA: Rancho Los Amigos Rehabilitation Engineering Center; 1980. p. 45–8.
17. Basmajian JV, Stecko G. The role of muscles in arch support of the foot. J Bone Joint Surg 1963;45A:1184–90.
18. Bontrager E. Footswitch stride analyzer. Bull Prosthet Res 1981;18(1):284–8.
19. DiVita P. The selection of a standard convention for analyzing gait data based on the analysis of relevant biomechanical factors. J Biomech 1994;27:501–8.
20. DiFabio RP. Reliability of computerized surface electromyography for determining the onset of muscle activity. Phys Ther 1987;67(1):43–8.
21. Perry J, Bontrager EL, Bogey RA, Gronley JK, Barnes LA. The Rancho EMG Analyzer: a computerized system for gait analysis. J Biomed Eng 1993;15:487–96.
22. Ounpuu S, DeLuca PA, Bell KJ, Davis RB. Using surface electrodes for the evaluation of the rectus femoris, vastus medialis and vastus lateralis in children with cerebral palsy. Gait Posture 1997;5:211–6.
23. Agarwal GC, Gottlieb GL. Mathematical modeling and simulation of the postural control loop, Part II. Crit Rev Biomed Eng 1984;11:113–54.
24. Corcos DM, Gottlieb GL, Latash ML, Almeida GL, Agarwal GC. Electromechanical delay: an experimental artifact. J Electromyogr Kinesiol 1992;2:59–68.
25. Ralston HJ, Todd FN, Inman VT. Comparison of electrical activity and duration of tension in the human rectus femoris muscle. Electromyogr Clin Neurophysiol 1976;16:277–86.
26. Vos EJ, Harlaar J, Schenau GJVI. Electromechanical delay during knee extensor contractions. Med Sci Sports 1998;23:1187–93.
27. Viitasalo JT, Komi PV. Interrelationships between electromyographic, mechanical, muscle structure and reflex time measurements in man. Acta Physiol Scand 1981;111:97–103.
28. Solomonow M, Baratta R, Shoji H, D'Ambrosia R. The EMG-force relationships of skeletal muscle; dependence on contraction rate, and motor units control strategy. Electromyogr Clin Neurophysiol 1990;30:141–52.
29. Lawrence JH, De Luca CJ. Myoelectric signal versus force relationship in different human muscles. J Appl Physiol 1983;54(6):1653–9.
30. Perry J, Bekey GA. EMG-force relationships in skeletal muscle. Crit Rev Biomed Eng 1981;7(1):1–22.
31. Perry J, Antonelli D, Ford W. Analysis of knee-joint forces during flexed-knee stance. J Bone Joint Surg 1975; 57A(7):961–7.
32. Smidt GL. Biomechanical analysis of knee flexion and extension. J Biomech 1973;6:79–92.
33. Horowits R, Podolsky RJ. The positional stability of thick filaments in activated skeletal muscle depends on sarcomere length: evidence for the role of titin filaments. J Cell Biol 1987;105:2217–23.
34. Hardt DE. Determining muscle forces in the leg during normal human walking: an application and evaluation of optimization methods. J Biomed Eng 1978;100:72–8.
35. Lyons K, Perry J, Gronley JK, Barnes L, Antonelli D. Timing and relative intensity of hip extensor and abductor muscle action during level and stair ambulation: an EMG study. Phys Ther 1983;63:1597–605.
36. Thomas SS, Moore C, Kelp-Lenane C, Norris C. Simulated gait patterns: the resulting effects on gait parameters, dynamic electromyography, joint moments and physiological cost indices. Gait Posture 1996;4:100–7.

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Dr. Perry joined the Rancho Los Amigos Hospital staff as the Chief of Orthopaedic Surgery of the Adult Poliomyelitis Service in 1955 and has been the Chief of Pathokinesiology Service, Director, Quality Assurance Program since that time. Paralytic hand dysfunction led her to initiate a dynamic electromyography program in 1961 to study hand muscle phasing with intramuscular (wire) electrodes. The same year, vaccine eradication of acute polio allowed Dr. Perry to start a CVA Service. Inconsistencies between observed gait deficits and clinical findings led to the initiation of a gait laboratory to define muscle function by EMG. A foot switch system was designed for timing. Motion analysis by video observation was later augmented by kinematic and kinetic systems. Technical refinements to allow clear delineation of adjacent muscle action and automated EMG interpretation have been her major objectives.

Dr. Perry has over 300 publications, the most prominent being her book *Gait Analysis: Normal and Pathological Function.* She has received 27 formal honors for her work in orthopaedic surgery, gait analysis, and rehabilitation.

In 1998, Dr. Perry and her staff moved into the Jacquelin Perry Neuro-Trauma Institute and Rehabilitation Center (JPI), a new three-story, state-of-the-art hospital. She was awarded the honorary degree of Doctor of Science by the University of California and the Helen J. Hislop award for Outstanding Contributions to Professional Literature at the 1998 APTA Annual National Conference Honors and Awards ceremony. Her current status is emeritus professor of orthopaedic surgery, emeritus professor of biokinesiology and physical therapy, medical consultant for the Rancho Los Amigos Pathokinesiology Service (which she established in 1968 for the primary purpose of studying normal and pathological gait) and Centinela Hospital biomechanics laboratory, Chief of the Post-Polio Service, and gait consultant to the Traumatic Brain Injury Service.

SECTION TWO

Chapter Two

Motion Analysis and Biomechanics

by Robert W. Soutas-Little, Ph.D.

Dr. Soutas-Little is a Professor of Theoretical Mechanics and Director of both the Biomechanics Evaluation Laboratory and Biodynamics Laboratory at Michigan State University in East Lansing, Michigan.

INTRODUCTION

Classical, or Newtonian, mechanics is the oldest branch of physics devoted to the study of motion, the forces that cause that motion, and the internal forces that act within the body. Biomechanics is the application of Newtonian mechanics to the study of the neuromuscular skeletal system. Biomechanics has found its greatest use in orthopaedics and physical medicine and rehabilitation characterizing function and dysfunction of the muscular skeletal system. One branch of biomechanics, gait analysis or motion analysis of human gait, has developed since early studies in the late 1900s. Motion analysis has been extended during the past two decades to investigate many other activities in addition to gait analysis. Currently, postural balance studies, stair ascending, or descending, and upper limbs are all being studied using motion analysis and the techniques of biodynamics. Although motion analysis requires the use of the mathematical techniques of dynamics, the presentation here will be on a conceptual basis where possible.

TEMPORAL PARAMETERS OF THE GAIT CYCLE

There are variations in the definitions of the different phases of the gait cycle during walking but the most commonly defined phases will be discussed here. The gait cycle is defined as the period from heel contact of one foot (for example, the left foot) to the next heel contact of the same foot. This cycle is broken into two parts, stance phase and swing phase. On the average, the gait cycle is about one second in duration with 60 percent in stance and 40 percent in swing. The stance phase is further divided into an initial double stance, followed by a period of single stance and then a final period of double stance. Double stance indicates that both feet are in contact with the ground; single stance is the period when only one foot is in contact with the ground. When walking, there must be a period of double stance and when running, this period is replaced by a flight phase during which neither foot is in contact with the ground. The walking gait cycle is illustrated in **Figure 1.** During the early part of stance phase, the heel is in contact with the ground, progressing to foot-flat during single stance and then to the forefoot contact during the final double stance phase ending with toe-off. This would be the normal contact areas of the plantar surface of the foot with the ground but may vary greatly with pathological gait. For example, equinus gait is characterized by the forefoot striking the ground first and then the contact area, progressing to the posterior in some cases while in others the heel never contacts the ground.

During double stance, the weight is transferred from one foot to the other. During single stance, the center of mass of the body passes over the foot in

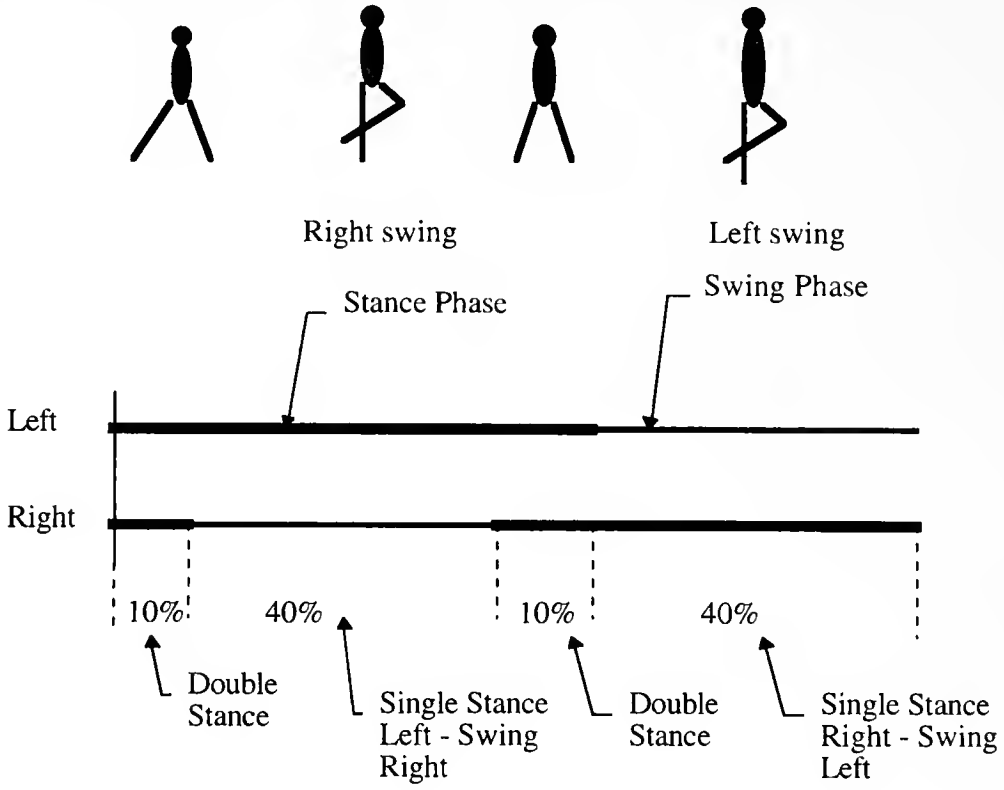

Figure 1 The Gait Cycle

Figure 1.
Temporal parameters of the gait cycle.

preparation for shifting to the other limb. Walking has been described as a series of falls from one limb to the other and it is obvious that the greatest danger of an actual fall is during this period of transferring weight.

BODY SEGMENTS

It is important to understand the basic assumptions that are made to analyze human motion using the techniques of rigid body dynamics. The most basic assumption is that body segments can be modeled as rigid bodies, that is, the position and motion of the underlying skeleton can be approximated by tracking the position and motion of the surface tissue. The error that arises is referred to as soft tissue motion error, which is inherent in all motion analysis of human subjects. Only a few research tests have been performed to test the amount of error due to soft tissue movement and these tests have involved invasive techniques of putting pins in the bones and attaching markers to those pins and comparing pin marker movement with that of surface markers (1). Therefore, it is important to place surface markers at points where soft tissue movement is a minimum. Obviously, this can present problems when testing individuals who are obese.

When a body is modeled as a rigid body, the distance between any two points on that body is constant. Consider a body segment such as the thigh modeled as a single rigid body, as shown in **Figure 2.** Three markers are shown on the body segment so that they are non-collinear, that is, they do not lie on a line. The markers form a triangle on the body segment and it is assumed that the lines AB, BC, and CA do not

change in length. The position of each marker is measured by a motion analysis system and this position is expressed as coordinates in a fixed laboratory reference system. Different systems use different orientation of the laboratory coordinate systems but the method of analysis is the same for all systems. Consider the laboratory system shown in **Figure 3** and the position vectors to the three segment markers. We will define the three position vectors as:

$$\begin{aligned} \mathbf{r}_A &= X_A(t)\,\hat{\mathbf{I}}+Y_A(t)\,\hat{\mathbf{J}}+Z_A(t)\hat{\mathbf{K}} \\ \mathbf{r}_B &= X_B(t)\,\hat{\mathbf{I}}+Y_B(t)\,\hat{\mathbf{J}}+Z_B(t)\hat{\mathbf{K}} \\ \mathbf{r}_C &= X_C(t)\,\hat{\mathbf{I}}+Y_C(t)\,\hat{\mathbf{J}}+Z_C(t)\hat{\mathbf{K}} \end{aligned} \qquad [1]$$

where $\hat{I},\hat{J},\hat{K}$ are the unit base vectors in the laboratory coordinate system, that is, they are vectors of magnitude one that serve as pointers in the *X*, *Y*, and *Z* directions, respectively. Note that the components of each position vector are the coordinates of the marker position and are shown as a function of time as the marker position will change as the marker moves. This position is measured at specified intervals in time and this interval is dictated by the camera speed. The camera speed is usually specified in Hz (Hertz) or pictures per second. Therefore, a 100 Hz system would take 100 pictures per second or at intervals of 10 ms. As previously stated, the vector multiplying each component of the position vectors is the unit base vector of the laboratory system and may be thought of as a pointer of magnitude one pointing in the coordinate direction. These unit vectors form the basis of all vector analysis and are fundamental to the understanding of biodynamics. The laboratory coordinate system is a right-handed coordinate system, that is, the *X* axis is aligned with the thumb of the right hand, the *Y* axis is aligned with the index finger of the right hand and the *Z* axis is aligned with the middle finger of the right hand. All coordinate systems used in vector analysis must be right-handed coordinate systems (See **Figure 4**).

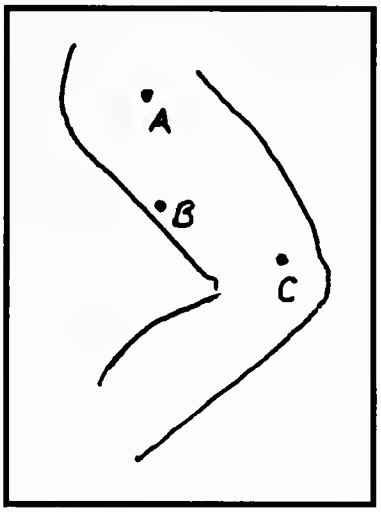

Figure 2.
Markers defining a rigid body segment.

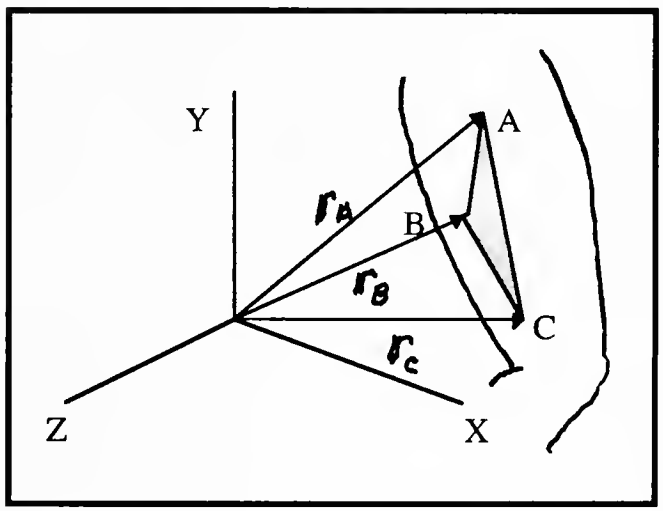

Figure 3.
Position vectors to body segment markers.

The position vectors to the three markers on the body segment will be used to obtain a segmental coordinate system, which may be thought of as three mutually perpendicular lines, attached to the body segment, that remain at a fixed orientation to that segment. In the discussion that follows, we will assume that the *x* segmental axis is in the anterior direction, the *y* segmental axis is in the medial-lateral direction directed to the left of the body segment, and the *z* segmental axis is directed in a superior direction on the body segment or directed distal to proximal in a lower limb segment. Although there are many different ways to form the segmental coordinate system, we will assume for this discussion that the three markers have been placed on the body segment such that two markers define a segmental axis and the three markers form a segmental anatomical plane. For example, on the thigh, markers A and C may define the superior axis of the thigh and the three markers are placed in a parasagittal plane of the thigh. A relative position vector from C to A is designated by $r_{A/C}$ (A relative to C) and is obtained

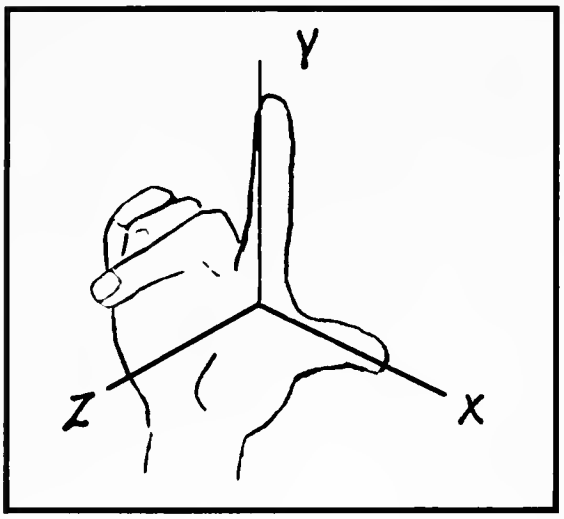

Figure 4.
Right hand coordinate system.

by subtracting the coordinates of marker C from those of marker A.

$$\mathbf{r}_{A/C} = \mathbf{r}_A - \mathbf{r}_C \quad [2]$$

It is important to realize that the length of this relative position vector does not change, since we have assumed that the body segment is rigid. The relative position vector may be thought of as a pencil glued to the body segment that is oriented with the body segment but does not change in length. If the length is computed on a frame-by-frame basis, the validity of the assumption may be measured. This length is called the magnitude of the relative position vector. A unit vector in the segmental coordinate direction z is obtained by dividing the relative position vector by its magnitude.

$$\hat{\mathbf{k}} = \frac{\mathbf{r}_{A/C}}{|\mathbf{r}_{A/C}|} \quad [3]$$

This unit vector will change its orientation in space but not its orientation relative to the body segment. A vector operation called the vector product, or cross product, is defined such that the resulting vector is perpendicular to the plane formed by two vectors. Let us form a second relative position vector from C to B

$$\mathbf{r}_{B/C} = \mathbf{r}_B - \mathbf{r}_C \quad [4]$$

If A, B, and C form a parasagittal plane, then a vector perpendicular to this plane in the medial-lateral direction can be obtained by taking the cross product between the relative position vectors defined in Equations 2 and 4 yielding a unit vector in the y segmental coordinate direction.

$$\hat{\mathbf{j}} = \frac{\mathbf{r}_{B/C} \times \mathbf{r}_{A/C}}{|\mathbf{r}_{B/C} \times \mathbf{r}_{A/C}|} \quad [5]$$

The final coordinate direction for the body segment is obtained by the cross product of the two segmental coordinate base vectors.

$$\hat{\mathbf{i}} = \hat{\mathbf{j}} \times \hat{\mathbf{k}} \quad [6]$$

The position of the body segment can now be determined by the position vector to marker C and the three-dimensional (3-D) orientation of the body segment is defined by the triad of segmental base unit vectors as shown in **Figure 5.** As mentioned earlier, there are many different protocols to place markers to define the segmental coordinate system but such a system must be formed for each body segment. Currently, there is no standard designation of which coordinate axis is oriented in which segmental direction. Here we have assumed that the x axis is in the anterior segment direction but some laboratories may designate the y or z axis in that direction. However, the segmental coordinate axes must be a right-handed coordinate system.

JOINT KINEMATICS

We have shown how the movement of a body segment can be tracked in the laboratory but what is of interest in the analysis of human movement is not the position and orientation (six degrees of freedom) of the

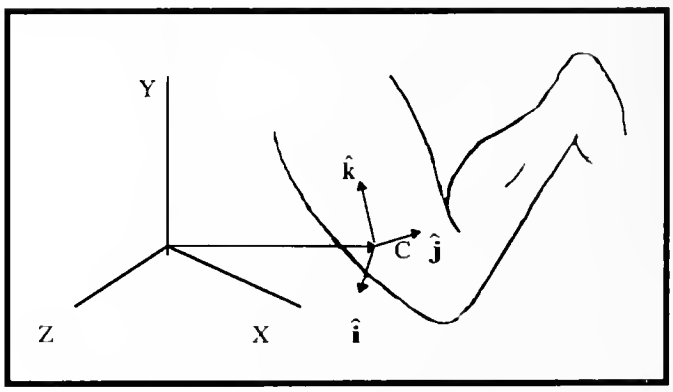

Figure 5.
Segmental coordinate system.

segment but the *relative* position and orientation of one body segment to the adjoining one; this describes the joint kinematics. The relative position of one body segment to another is easier to obtain but more difficult to interpret clinically so it will be discussed later. The relative orientation of one body segment to another defines the joint angles. For example, the orientation of the tibia to the femur defines the three clinical angles of flexion/extension, abduction/adduction, and internal/external rotation of the tibia relative to the femur.

Before we can define these clinical angles, we must consider finite 3-D rotations in general. It has been recognized for over 200 years that 3-D rotations are sequence dependent. This is illustrated in **Figure 6;** rotate a book 90° first about the x-axis (a)→(b) and then 90° about the y-axis (b)→(c). The order of rotations is then reversed: first rotate 90° about the y-axis (d)→(e) followed by a 90° rotation about the x-axis (e)→(f). This problem was first addressed by the Swiss mathematician Euler in 1776 (2). He recommended that rotations be defined by a sequence of three rotations. We will examine the knee joint angles defined by a sequence of rotations of the tibia relative to the femur using a set of Euler angles suggested by Grood and Suntay (3) in 1983. We will represent the femoral coordinates with capital letters and the tibial coordinates with lower case letters. We will rotate first about the Y axis of the femur yielding the flexion/extension angle. The tibial x and z axes will no longer be parallel with the femoral X and Z axes and the tibial axes will be designated by x′ and z′. The second rotation will be about the current tibial x′ axis yielding the abduction/adduction angle. The tibial y′ and z′ axes will have rotated to a new position designated by y″ and z″. The final rotation will be about the tibial z″ axis yielding internal/external tibial rotation. These rotations are illustrated in **Figure 7.** Note that the first rotation was about a femoral axis and the last rotation was about a tibial axis. The intermediate rotation is not about a current femoral or tibial axis but about an intermediate tibial axis. This is denoted as a Yxz rotation sequence, that is, rotation about the Y axis of the femur, followed by rotation about the intermediate x axis of the tibia and finally rotation about the z axis of the tibia. Euler (3) called this second axis the *line of nodes* and Grood and Suntay called it *a floating axis* (2). All joint analyses are

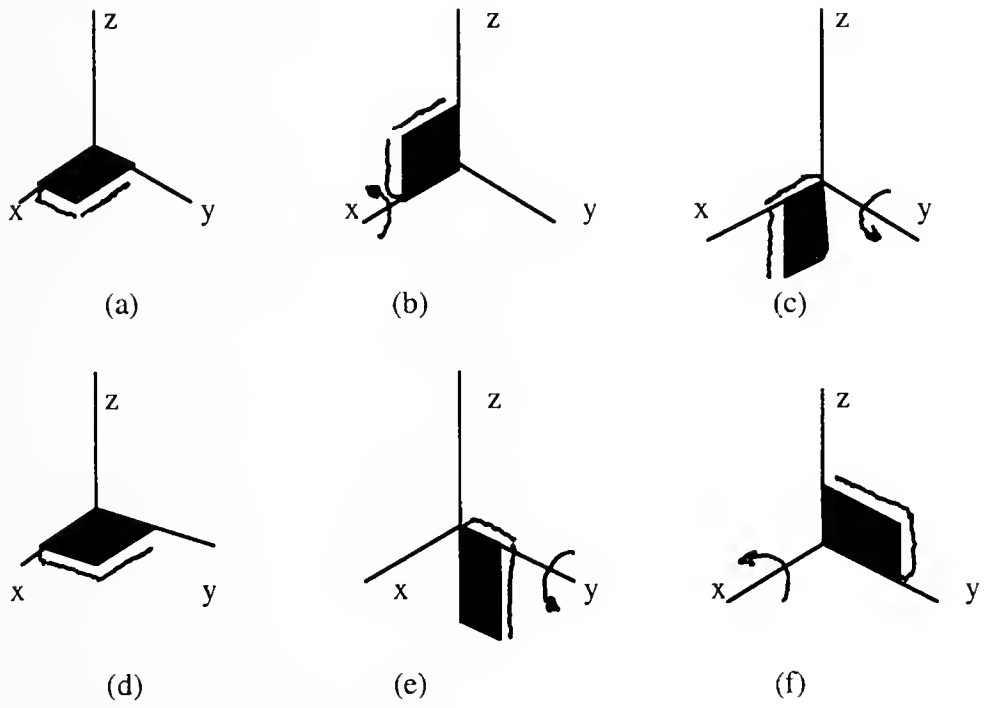

Figure 6.
Sequence dependence of finite rotations.

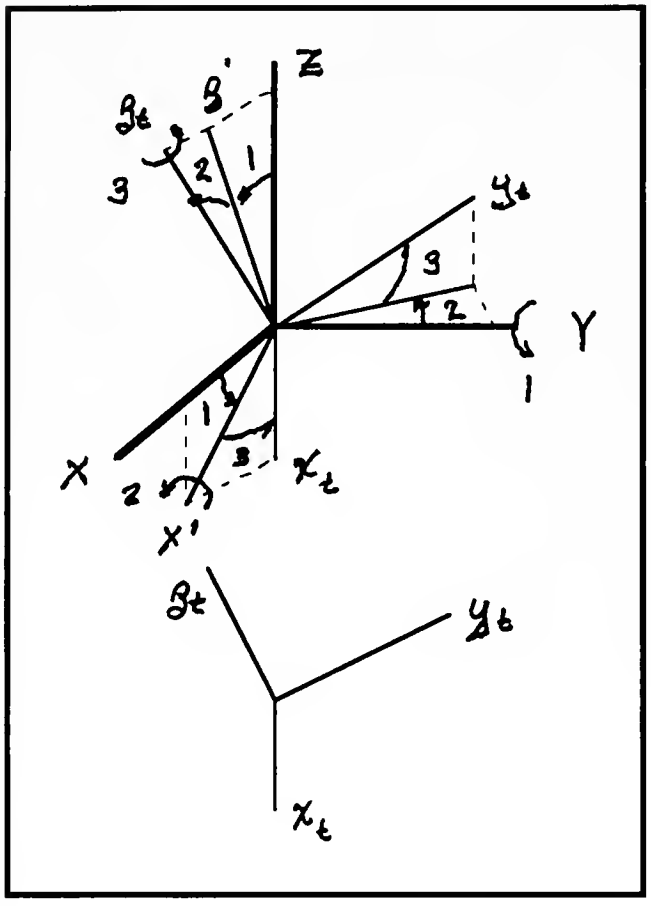

Figure 7.
Joint coordinate system.

based on a form of Euler angles. The difficulty is that different laboratory or motion analysis system manufacturers define the sequence of rotations in different orders, resulting in different joint angles. Care should be taken when comparison of data from one laboratory to another is made or in the interpretation of published data.

An example of the clinical importance of the sequence of rotations can be illustrated by examining head position relative to the thoracic cage. We will take the thoracic cage as the stationary or reference segment and examine head position relative to it. Examine two different sequences of rotations for the clinical interpretations of the motions. We will first do a (Yxz) sequence as was defined for the knee. Head flexion/extension will be defined as a rotation about the medial-lateral axis on the thoracic cage (the Y axis), followed by a rotation about the posterior-anterior head axis at this intermediate position to obtain lateral side bending of the head and finally rotation about and inferior-superior head axis to define head rotation left or right.

Mathematically, we could alternately define the head position as a (Zxy) sequence. Head rotation left and right will first be defined by rotation about the vertical axis of the thoracic cage, followed by lateral side bending about the intermediate position of the anterior axis of the head, and finally, flexion/extension will be defined as a rotation about the medial lateral axis of the head. If you examine these two different definitions of the head angles, you will see that they are not only different in magnitude but in clinical interpretation. The question arises whether head flexion should be defined as a rotation about the medial-lateral axis of the thoracic cage or a rotation about the medial-lateral axis of the head regardless of the position of the head. It should be stressed that both are mathematically correct so that either is acceptable to the physical scientist. The clinical interpretation is whether cervical flexion and extension is predominantly motion in the lower or upper cervical spine segments. The first sequence argues that flexion/extension are primarily motions in the lower cervical spine and rotations left and right are motions in the higher cervical spine. The second sequence argues the reverse.

A more complex example of 3-D coupled motions is the shoulder joint. At the present time, only limited investigations have been made of this joint's motion and most of these have been cadaveric studies. Investigators at Mayo Clinic suggested a Zxz transformation sequence, that is, first circumduction about the vertical axis of the thoracic cage, followed by flexion about the current humeral axis (the floating axis), and finally internally or externally rotating about the distal/proximal axis of the humerus. This sequence of rotation was the one originally proposed by Euler in 1776 to describe the motion of a spinning top.

It is felt by most biomechanists that the choice of joint axes should be made on a joint-by-joint basis in order to give the most relevant clinical information. There have been attempts during the last few years to establish standard definitions but to date none have been established.

It is important to note that the coupling of the three joint rotations is very important clinically. Foot motion is frequently defined as pronation or supination, that is, a coupled motion of the ankle involving dorsiflexion, eversion, and medial rotation during pronation and

plantar flexion, inversion, and lateral rotation during supination. The motion of almost all joints cannot be described by simple two-dimensional (2-D) definitions.

Joint Center

If we are to discuss the moments or torques acting on a joint, we must define a point within the joint known as the joint center. For some joints, such as the hip, this point is easily defined from anatomical considerations. The hip joint is modeled as a ball and socket joint or, more precisely, a ball and half-socket joint. Therefore, the joint center is taken as the center of the spherical femoral head. The femur rotates about this point in movements of the femur relative to the pelvis and this point is stationary on both the pelvis and the femur. For other joints, there is not a clearly defined joint center. For example, the knee joint is characterized by the femur both sliding and rolling on the tibia and there is no single point that acts as a hinge point. We are then faced with the problem of defining an equivalent joint center. The easiest way is to define the geometric center of the joint as the joint center and this is set to be equal to the midpoint between the femoral condyles. Although this is certainly not the kinematic or rotational joint center, it will provide a reproducible reference point for the analysis of the joint moments.

It is possible to define a kinematic joint center using an *instantaneous center of rotation* for sagittal plane analysis or an *instantaneous helical axis* for general 3-D analysis. It is beyond the scope of this Chapter to discuss the mathematics to obtain these centers but the concept will be briefly presented. In a 2-D analysis of the motion of one rigid body relative to another, at any instant of time the two bodies rotate such that they appear to be hinged at a single point. At this instant, the velocity of this point will be the same if the point is assumed to be on the first or second rigid body. To understand this concept, we must first look at the linear velocity of a point on the rigid body and the angular velocity of the body.

Angular Velocity of a Body Segment

Consider two points on a body segment that is modeled as a rigid body as shown in **Figure 8.** The relative position vector $r_{B/A}$ cannot change in length nor move on the body segment but may change orientation in space as the body segment rotates. In fact, this is how the rotation of the body is tracked. Points A and B have absolute linear velocities in space that are not, in general, equal. The linear velocity of a point is defined

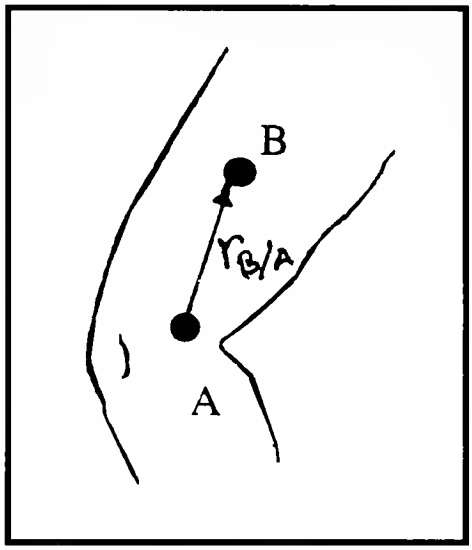

Figure 8.
Relative position vector.

as the time rate of change of the position vector to that point and the velocity is defined as the derivative of the position vector with respect to time as shown in **Figure 9.**

If we let $\Delta t=t'-t$, then the velocity is defined as

$$\mathbf{v} = Lim_{\Delta t \to 0} \frac{\Delta \mathbf{r}}{\Delta t} = \frac{d\mathbf{r}}{dt} \qquad [7]$$

The relative velocity of point B to point A in **Figure 8** is defined as:

$$\mathbf{v}_{B/A} = \mathbf{v}_B - \mathbf{v}_A \qquad [8]$$

Since the length of the relative position vector of B relative to A cannot change in length, the only thing that B may do relative to A is to rotate about it. This is fundamental to the definition of rotation of a rigid body or, in this case, a body segment. This relationship is expressed mathematically as:

$$\mathbf{v}_{B/A} = \omega \times \mathbf{r}_{B/A} \qquad [9]$$

where ω is the angular velocity of the body measured in radians per second. (2π radians=360°). The rigid body is said to have an angular velocity of ω at this instant of time. The angular velocity is a vector having both a magnitude and a direction. The orientation of the vector is the axis of rotation and the sense is given by a right-hand rule, that is, if the thumb of the right hand is pointed in the direction of the vector, the fingers curl

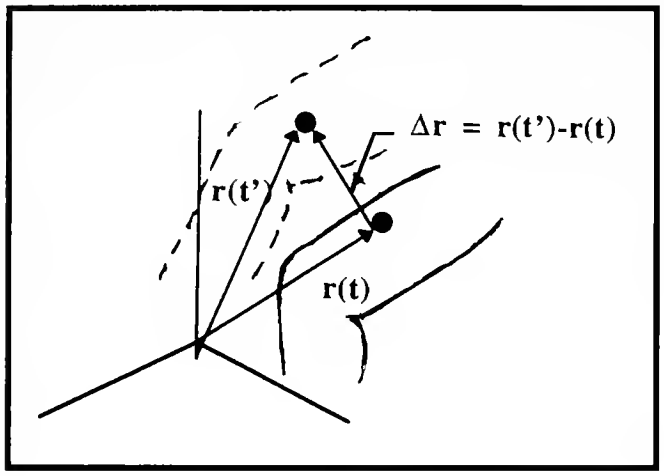

Figure 9.
Change in a position vector.

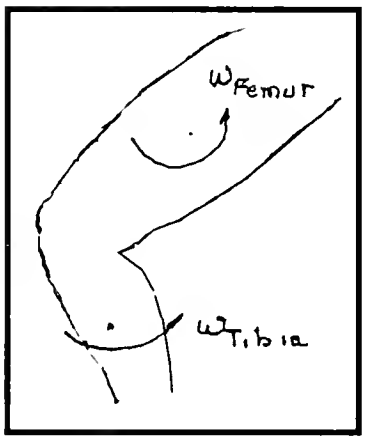

Figure 10.
Angular velocity of the knee joint.

around the vector and designate the rotation. The angular velocity vector designates the rotation of the body segment in an absolute sense (i.e., relative to the fixed laboratory coordinates).

The angular velocity of the joint is the relative angular velocity of the body segment distal to the joint relative to the proximal segment. Therefore, the angular velocity of the knee is:

$$\omega_{Knee} = \omega_{Tibia} - \omega_{Femur} \qquad [10]$$

This is illustrated in **Figure 10.**

The angular velocity of the joint will have components in three directions and these components

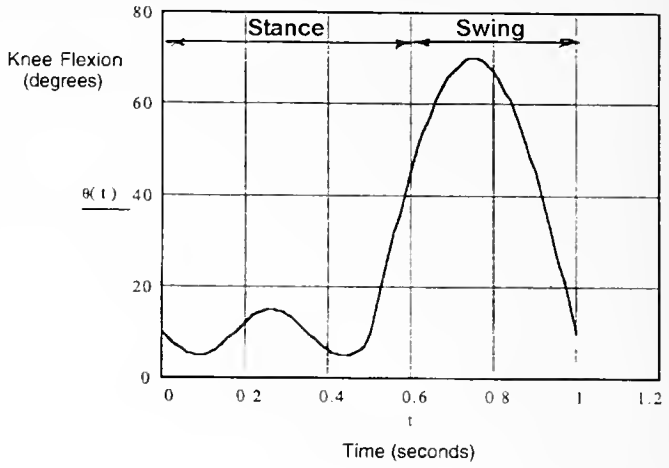

Figure 11.
Knee flexion for walking gait cycle.

will be equal to the rate of flexion/extension, abduction/adduction, and internal/external rotation. It is important to realize that the rate of change of a joint angle is different from the joint angle. For example, the knee may be flexed at a given time but be flexing or extending at the same time. If curve is plotted of the knee flexion/extension, the slope of this curve is the angular velocity component in flexion and extension. A typical flexion/extension curve for the knee is shown in **Figure 11;** the knee is approximately 10° flexed at heel contact and extends for the first 100 ms. This is followed by flexing and extending during single stance phase followed by flexing through toe-off to midswing. The final phase of swing is characterized by knee extension from the maximum flexed position. Note that the angular velocity of the knee in flexion and extension is the slope of this curve. If the slope is positive, the knee is flexing and if the slope is negative, the knee is extending. The maximum angular velocity of the knee occurs both during initial swing and late swing.

The concept of an instantaneous center of rotation or hinge point is a point on the femur that coincides with a point on the tibia that has the same velocity. When this concept is extended to three dimensions, one rigid body appears, at any instant, to rotate about an axis in space and to slide along that axis relative to the other body; hence, the term "instantaneous helical axis" or "screw axis." Successive helical axes will intersect at points relative to the femur and the centroid of these intersections is defined as the joint center as shown in **Figure 12.** It is important to compare data

only as the definition of a particular joint center is the same in each analysis.

GROUND REACTION FORCES (GRF)

During gait, when the foot is in contact with the ground it applies a force to the ground and a GRF is developed that is equal and opposite to the force the foot applies on the ground. We are interested in this GRF because this is an external force acting on the body while walking. The only other external force acting on the body is gravitational attraction if wind resistance or drag is neglected. The force the foot applies to the ground is measured by a force plate or a dynamometer that is mounted securely in the floor such that its surface is flush with the floor (see **Figure 13**). The force plate has an instrument center that is below the floor and the resultant force and moment about this instrument center is measured. These data are sampled at a specific rate, usually 1000 Hz, or every millisecond. The resultant force and moment are expressed in an equivalent force system composed of the resultant force acting at a specific point on the surface of the force plate and a torque about the vertical axis. The resultant ground-reaction force is divided into three components: vertical, anterior/posterior, and medial/lateral. The torque is called the ground reaction torque and the unique point of the intercept of the GRF with the force plate surface is called the center of pressure (COP) or the center of force. The COP changes during stance phase generally moving from the rear of the foot anterior toward a point between the first and second metatarsal heads. The path of the COP on the force plate can be related to the path of the resultant ground-reaction force on the plantar surface of the foot. If an actual pressure distribution plot were obtained at an instant during stance phase, the COP would be the centroid of the pressure distribution. The COP path is also generated in this manner when pressure mats are used. Both pressure mats and force plates have been discussed in the Introduction.

Let us consider each component of the GRF separately; the largest is the vertical component and accounts for the acceleration of the body's center of mass in the vertical direction during walking. A typical plot of the vertical ground-reaction force is shown in **Figure 14** where the vertical reaction force is expressed in percent of body weight (%BW). This curve is sometimes called the **M** curve because it resembles that

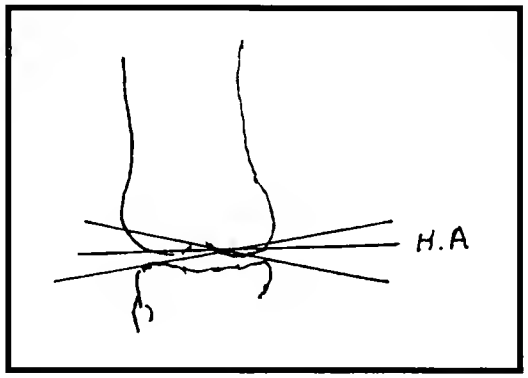

Figure 12.
Intercepts of helical axes (HA).

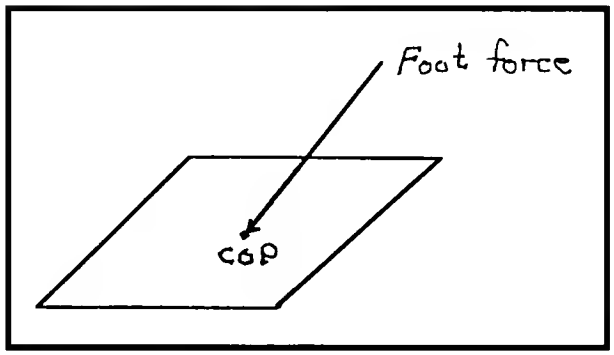

Figure 13.
Force plate and center of pressure (COP).

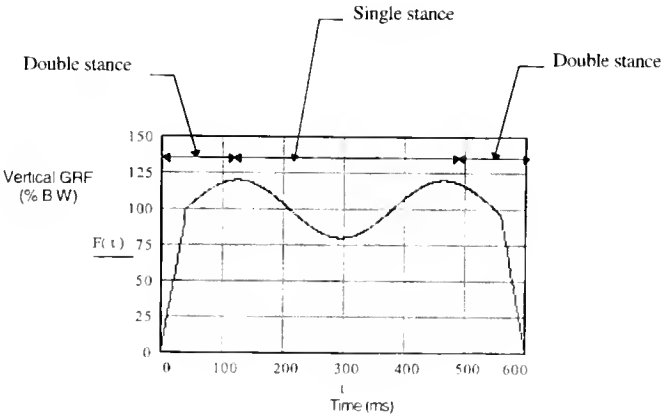

Figure 14.
Vertical ground reaction force (GRF) during walking.

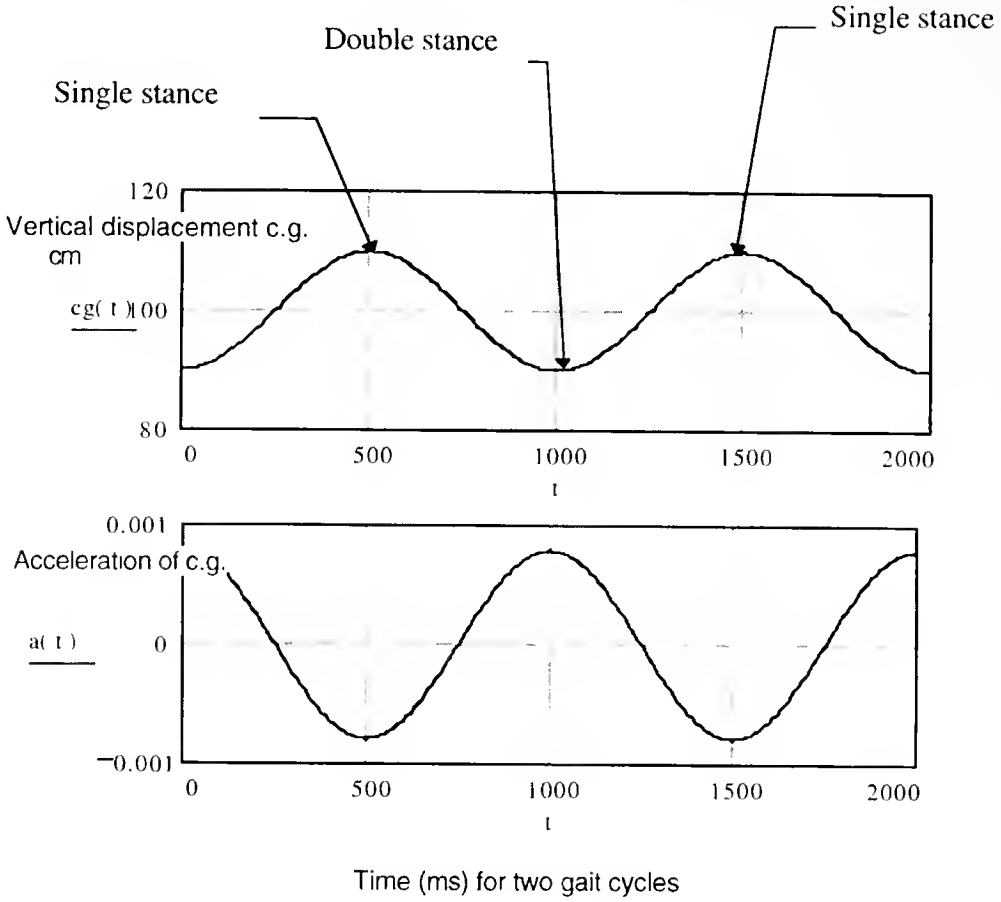

Figure 15.
Vertical displacement and acceleration of the center of gravity (c.g.) of the body during walking.

letter. During the first 100 ms, the GRF goes to a maximum of 120%BW during the double stance phase. During single stance phase, the vertical GRF drops to about 80%BW or for the more dynamic walker to 60 to 70%BW. At first, it seems unusual that the GRF should be less than body weight during single stance when only one foot is on the ground. This is made clearer if the vertical position of the center of mass of the body during the gait cycle is examined. The center of mass is located around the center of the pelvis, ignoring changes due to arm position, and executes a sinusoidal motion rising and falling about 10 cm in space during walking, as shown in **Figure 15.** The acceleration of the center of mass in the vertical direction is shown below the displacement of the center of mass and it can be seen that this is opposite in sign at each point in the gaitcycle. If the entire body is treated as a mass on a spring, the magnitude of the GRF can be more easily understood. In **Figure 16,** the body is shown as a single mass and indicates the forces acting on the mass. Newton's second law states that the unbalanced force must equal the mass times the acceleration. Therefore, when the acceleration is positive, the GRF must be greater than BW. The positive acceleration occurs during double stance when the center of mass is at its lowest point. When the center of mass is at its highest point during single stance, the acceleration is negative and GRF must be less than BW. During a more dynamic gait, the vertical excursion of the center of mass is greater and vertical GRF will have a greater deviation from the BW. The GRF in the elderly remains at approximately BW during single stance.

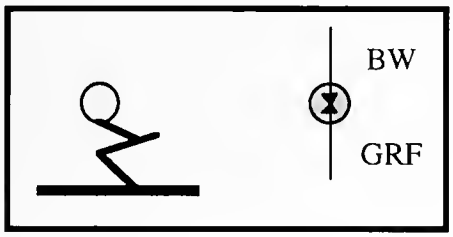

Figure 16.
Particle model of the body.

The anterior-posterior (AP) GRF is first a braking force to mid-stance, followed by propulsion, and usually represents a sine curve with an amplitude of 25%BW, as shown in **Figure 17.** The AP GRF is braking for approximately 50 percent of stance phase followed by propulsion. The area under any segment of this curve represents the impulse or the time integral of the force. The braking impulse should be approximately equal to the propulsion impulse for balanced gait left to right. The total impulse in the AP direction for a full gait cycle should be zero, as the impulse is equal to the change in momentum in the forward direction. If the individual is walking at a constant speed, there is no change in momentum and, therefore, no net impulse. If there is greater propulsion impulse on the left side as compared with the right and greater braking impulse on the right as compared with the left, the net impulse for the complete gait cycle can still be zero. However, in this case, greater demands are being placed on the left leg to maintain a constant speed. This is seen frequently in cross-country runners after a stress fracture. The runner will be rehabilitated for a stress fracture on the right leg and returned to competition. However, the runner will be unbalanced in function, placing higher demands on the left leg. If this state is allowed to continue, it usually results in a stress fracture of the uninjured leg due to the higher functional demands placed on that leg. All cases of this nature should be tested on the force plate and balanced by a trainer when necessary. The unbalanced AP force is compared with normal as shown in **Figure 18.**

The medial-lateral force is of lower magnitude in most situations and relates to balance during walking. The medial-lateral GRF initially acts in the medial direction with a magnitude of 10%BW or less and then acts laterally during the balance of stance phase.

The vertical ground reaction torque has received much less attention in gait analysis but is felt to be involved in the pelvic twist and arm swing during gait. It also has been used as a balance measurement in postural balance studies.

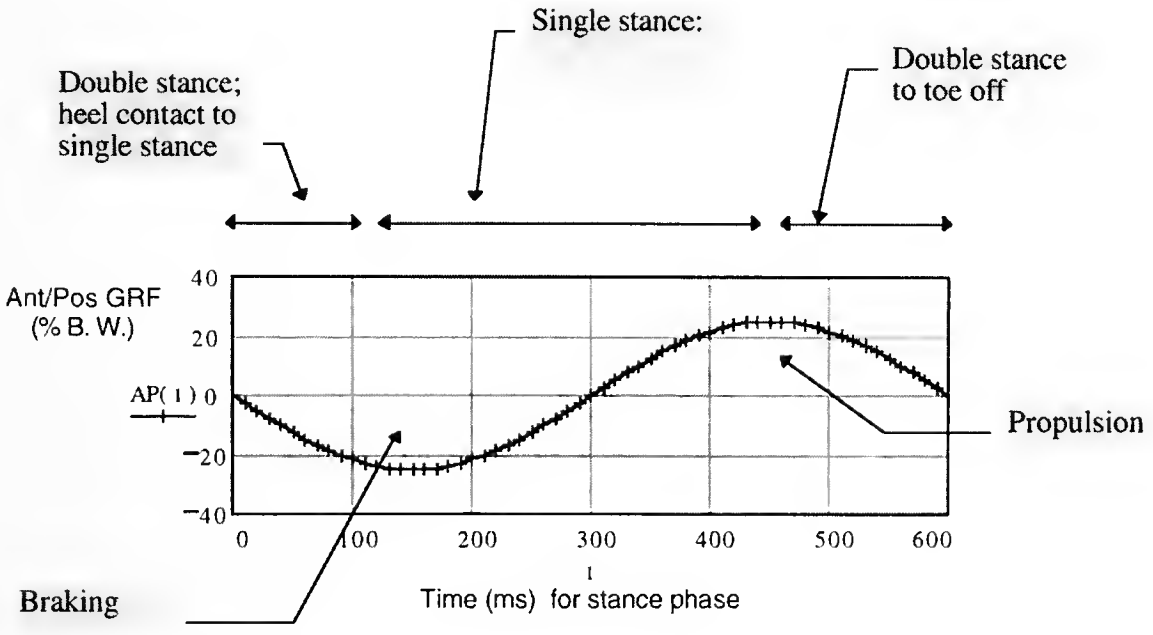

Figure 17.
Anterior-posterior (AP) ground reaction force (GRF).

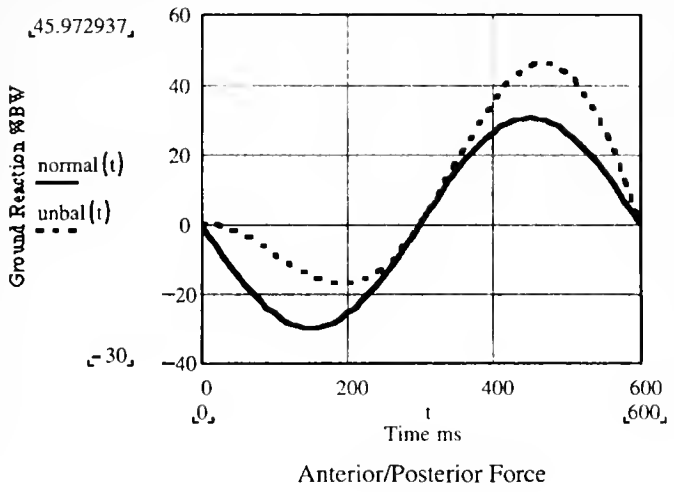

Figure 18.
Anterior-posterior ground reaction force for normal subject and subject with unbalanced limb.

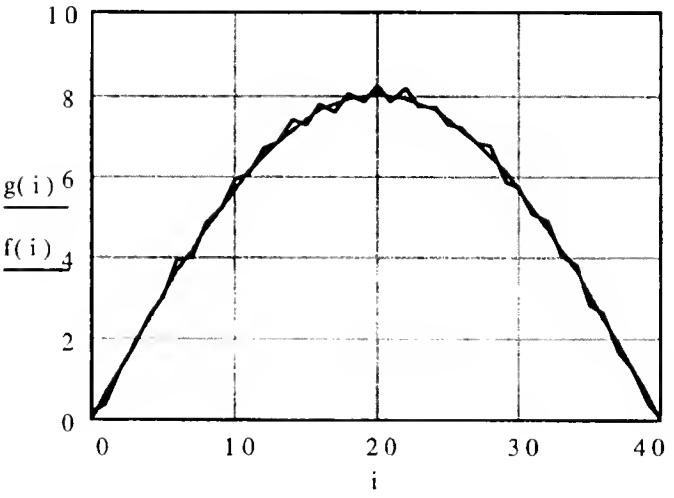

Figure 19.
Noise in position data.

KINETIC STUDIES

In most engineering studies of dynamics, the direct dynamics problem arises, that is, the forces are known and are used to develop the differential equations of motion. These equations are usually nonlinear in nature and numerical solutions are sought. The solution involves integration of the equation; a process, which by its nature, smoothes out noise in the force data. On the other hand, most biodynamics problems involve the inverse dynamics solution, wherein the motion of the body is known and the forces are obtained by differentiation of the position-time curves. Mathematically, this is an easier process if the position is known as a continuous function of time. However, as has been discussed in the Instrumented Gait Analysis Section and also in this Chapter, motion data are not obtained as continuous functions of time but at discrete intervals depending upon the speed of the video cameras. The forces and moments are then obtained by differentiation of these data. Since analytical differentiation cannot be used, the data must be differentiated numerically.

For an example of the inverse dynamics problem, consider that the displacement given is the function of time as:

$$x(t) = \sin(\pi t)\, e^{t}.$$

The velocity and acceleration are obtained by successive differentiation of this function:

$$v(t) = e^{t}\,[\sin(\pi t) + \pi \cos(\pi t)]$$
$$a(t) = e^{t}\,[2\pi \cos(\pi t) + (1 - \pi^{2}) \sin(\pi t)].$$

In this case, the position was given analytically as a continuous function of time.

Modern motion analysis equipment allows measurement of position data from 50 to 200 times a second (frame rates 50–200 Hz) using high-speed video cameras. This means that the position data are not known as a continuous analytical function but at discrete times. Higher frame rates can be obtained for use in measurements of high velocity movements. As previously noted, most experimental systems do not collect position data as a continuous function of time but at specific intervals of time. The velocity and acceleration are obtained by numerical differentiation of these data and are thus subject to increased noise in the calculation of the velocity and acceleration. As an example, suppose that the correct position function is

$$g(t) = 8\left[\sin\frac{\pi t}{40}\right]$$

and the position datum f(t) has noise of a random nature and maximum magnitude of 0.3 units as shown in **Figure 19.** These are not actual data but a file created using a random number function. Now we numerically differentiate both g(t) and f(t) to see the effect of the noise on the differential. The differentiated functions dg(t) and df(t) are shown in **Figure 20.** It is clearly seen in this file that the differentiated data are too noisy to be reliable. To completely describe the motion, the data

would have to be differentiated a second time to obtain acceleration. In actual practice, the position data are filtered using digital filters to smooth the data and to obtain more reliable differentiation in the presence of noise. Numerical differentiation techniques and digital filters are beyond the scope of this Chapter but the user of any motion analysis system should be aware of what types of differentiation and filtering routines are used.

Once the data have been filtered and differentiated, the joint forces may be obtained by solving from the most distal segment proximally. Each segment is modeled as a rigid body and isolated from the other segments but shows all forces acting on the segment—a free-body diagram; an example is shown in **Figure 21.** In this case, the GRF and the segment weight (W) are known and the ankle joint force (JF) and moment (Ma) are sought. The Euler-Newton equations of motion for a rigid body are:

$$\sum \mathbf{F} = \frac{d\mathbf{p}}{dt}$$
$$\sum \mathbf{M}_{\text{c.m.}} = \frac{d\mathbf{H}_{\text{c.m.}}}{dt} \qquad [11]$$

where **p** is the linear momentum of the segment and is equal to m**v**, that is, the product of the mass of the segment and the velocity of the center of mass of the segment and where $H_{c.m.}$ is the angular momentum of the segment about the center of mass. The angular momentum is equal to:

$$\mathbf{H}_{\text{c.m.}} = I_{xx}\omega_x \hat{\mathbf{i}} + I_{yy}\omega_y \hat{\mathbf{j}} + I_{zz}\omega_z \hat{\mathbf{k}} \qquad [12]$$

where the I's are the mass moments of inertia about the segmental coordinate axes. The mass moment of inertia is a measure of resistance to the angular acceleration about each coordinate axis. Values of the mass, location of the center of mass and the mass moments of inertia are available in the literature. The right side of Equation 11 is known and these algebraic equations can be solved for the joint force and moment. The proximal joint force and moment on the next is solved in a similar manner now that the distal joint force and moment for this segment are known.

The Euler-Newton equations of motion given in Equation 11 can be expanded into six scalar equations. Three equations for the linear momentum yielding:

$$\sum F_x = ma_x$$
$$\sum F_y = ma_y \qquad [13]$$
$$\sum F = ma_z$$

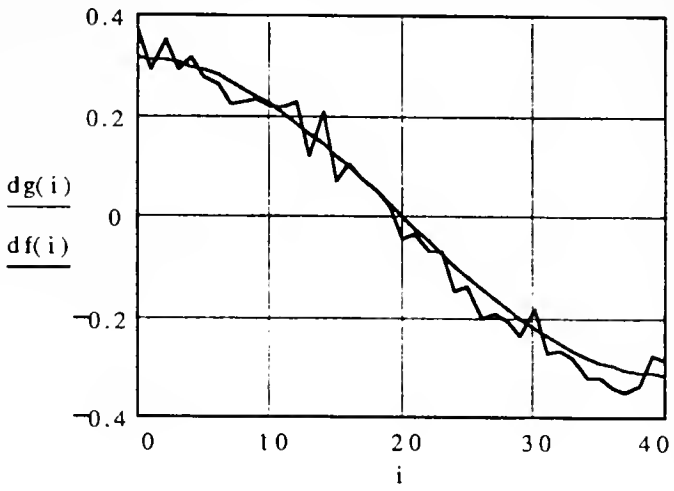

Figure 20.
Noise in velocity data.

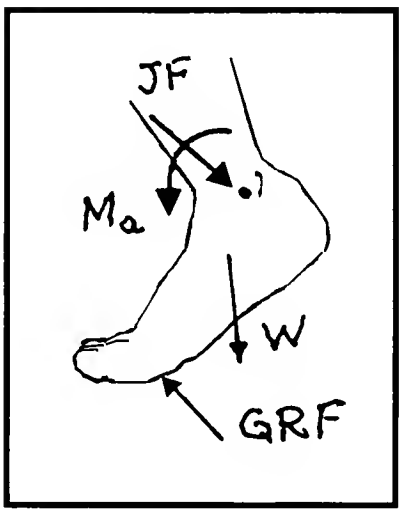

Figure 21.
Free-body diagram of foot.

The three equations for the angular momentum are more complex, yielding:

$$\sum M_x = I_{xx}\alpha_x - \omega_y\omega_z(I_{yy} - I_{zz})$$
$$\sum M_y = I_{yy}\alpha_y - \omega_z\omega_x(I_{zz} - I_{xx}) \qquad [14]$$
$$\sum M_z = I_{zz}\alpha_z - \omega_x\omega_y(I_{xx} - I_{yy})$$

The right-hand side of Equations 13 and 14 are known from motion analysis data and the forces and moments

at the proximal joint of the segment are determined by solution of these simultaneous equations.

Quasi-static Determination of the Joint Moments

In many cases, during slow walking, stair ascending or descending, or mild squatting, sufficient accuracy during stance phase can be obtained by ignoring the right-hand side of Equations 13 and 14. These terms are called the inertial forces, or inertia terms, and depend upon the linear and angular acceleration of the body segments. Especially during stance phase of gait, these terms are small compared to the GRFs, and the joint forces are calculated from the GRFs only. We may get a conceptual idea of the joint moments by examining **Figure 22.**

Figure 22 is not meant to be an exact representation of the location of the joint centers or the GRFs at some moment during gait but will give an example as how quasi-static moments can be determined. Although there are formal methods to determine the moments at any joint center using vector algebra and are used in most clinical software packages, the concept of sagittal plane joint moments may be obtained from this simple figure. The GRF has been shown as two separate components, one acting in the anterior direction and the second in the vertical direction. The simplest definition of a moment is the magnitude of the force times the perpendicular distance from the joint center to the force, that is:

$$M = F\,d \qquad [15]$$

Examining the ankle joint represented by the ankle joint center, we can see that both the anterior GRF and the vertical reaction force produce an applied or external dorsiflexion moment. If the ankle is not to collapse, the muscle moment must be equal and opposite to the applied moment. Therefore, there must be a net plantar flexion muscle moment or the gastrosoleous complex must produce the dominant muscle activity.

At the knee joint, the vertical GRF is a knee *flexion* force while the anterior GRF would cause knee *extension*. It is not obvious which of these applied moments will dominate since the vertical GRF is larger but has a lower moment arm (perpendicular distance from the joint) while the lower anterior GRF has a greater moment arm.

At the hip joint, both the anterior and vertical GRFs produce applied flexion moments, which must be resisted by the hip extensors. At first, this may appear to be a simple method to determine the dominant muscle activity at any joint. However, this is compromised by the fact that many muscles are two joint muscles producing opposite reactions at each joint (e.g., the hamstrings are hip extensors and knee flexors). This does give a feeling of the antagonistic muscle activity.

A similar view of the moments in the frontal plane can be obtained by examining **Figure 23;** wherein the net GRF passes medial to the knee center and causes an applied knee adduction moment during all of the stance phase of gait. If the individual is suffering from medial knee pain, each step aggravates this condition. If this moment is the cause of the pain, it is easy to determine by asking the individual to rotate his/her foot laterally, thus reducing the moment arm of the force from the joint center.

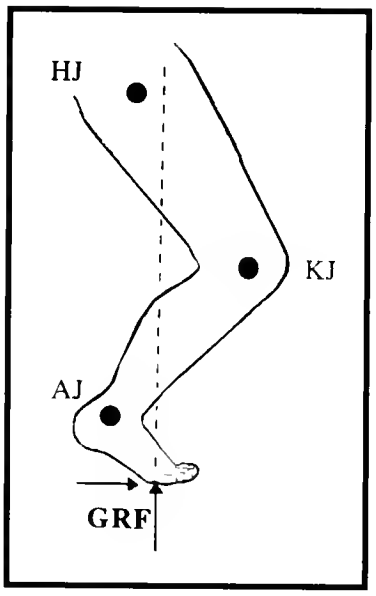

Figure 22.
Joint centers and line of action of ground reaction forces (GRF). AJ (ankle joint), KJ (knee joint), and HJ (hip joint).

Body Segments Considered as Levers

The easiest way to begin to understand the forces produced by the muscles spanning a joint and, therefore, the loads that are applied to the joints, is to consider simple levers, as shown in **Figure 24.** For this system to be in equilibrium (static balance), the total force acting on the lever must be zero and the turning effect, or the moment of the forces about the fulcrum *C* must be zero. These two conditions may be written as:

$$F_C = F_A + F_B$$
$$F_A a = F_B b \qquad [16]$$

Note that since the moment arm to F_b is greater than the moment arm to F_a, the force at A must be bigger than the force at B to balance the moments or the turning effects of the forces. This can affect the loads on the spine when an individual is lifting an object. The fulcrum point of the lever model of a lifting motion is the spine, in this case a point on the lumbar spine, and a lever model is shown in **Figure 25** One can take some typical values of the weights lifted and the weight of the torso and compute the force that the posterior back muscles must resist. If the individual is lifting 50 lb and the upper portion of the body is 120 lb, the moment arms to the weight, the center of mass of the torso, and to the back muscles, may be estimated as:

$$d_L = 20 \text{ in}$$
$$d_T = 15 \text{ in}$$
$$d_m = 1 \text{ in}$$

The muscle force is the only force resisting the tipping of the body about the fulcrum point on the lumbar spine. Therefore, balancing the moments about the spine yields:

$$M(1) = 50(20)+120(15) = 2800 \text{ lb}$$

At first glance, this may seem to be an impossible load as it would indicate that the compression on the spine would be the sum of the weight lifted, the weight of the torso, and the muscle force, or combining to be almost 3,000 lb. Because of the short moment arm to the muscles, these loads are correct and have been verified by experiments. Very few people realize the loads that are placed on the skeletal system due to the fact the muscles get "the short end of the stick."

The forearm provides another excellent example of this effect. Consider the forearm modeled as a simple lever as shown in **Figure 26**. A simple measurement on one's forearm will show that the length from the elbow to the palm of the hand is 8 to 10 times longer than the length from the elbow to the tendon attachment of the biceps. This means that if 25 lb are held in the palm of the hand, the muscle force must be between 200 and 250 lb. The compression on the elbow joint **C=M−W** would be between 175 and 225 lb. It is easily seen that if the lever arm to the weight were increased with a tennis racket or a shovel, the compression on the joint would increase even more.

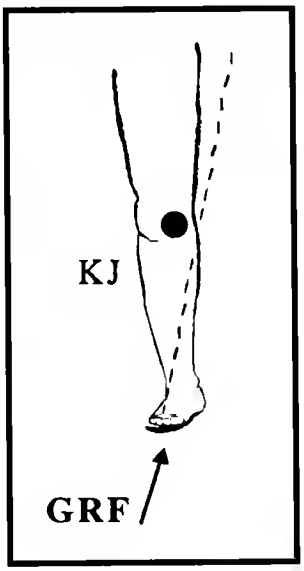

Figure 23.
Knee joint center and line of action of ground reaction force (GRF).

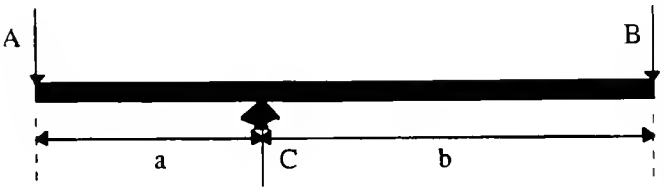

Figure 24.
Simple lever.

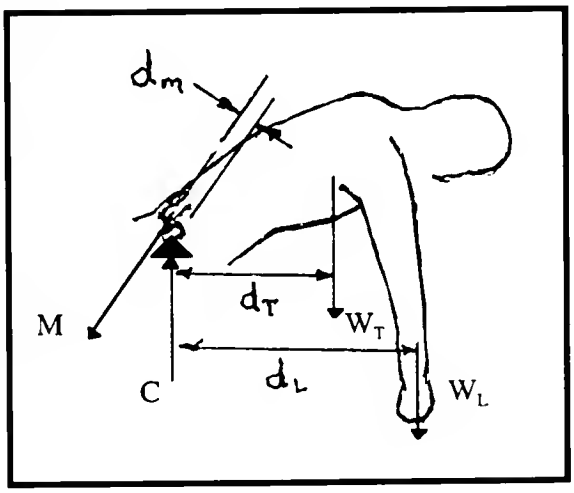

Figure 25
Lever model to determine lumbar spine load.

Muscle attachments are close to the joint centers so that small contractions of the muscle can produce large movements at the end of the levers producing a mechanical advantage in the motion of the limbs. Almost all loads on the muscular skeletal system can be modeled and understood by simple levers. These concepts can be applied in exercise therapy, orthotics, and in the understanding of the causes of injury.

POWER

Examination of the power expended by the muscles during a particular activity is a new tool that is being applied to gait analysis. Power is a measure of the rate of doing work, which is the product of the force applied in a certain direction and the distance the object moves in that direction.

$$\text{Work} = F\ d \qquad [17]$$

where F is the force and d is the distance the object moves in the direction of the force.

The power expended is the rate at which work is performed or the product of the force applied in a certain direction and the velocity of movement in that direction.

$$P = F\ v \qquad [18]$$

where v is the velocity in the direction of the force. Power is measured in Watts or Newton meters per second (N m/s). A moment also does work when it rotates an object through an angle, and the power performed by the moment is the product of the moment and the angular velocity of the object. Consider the power expended by the biceps when the elbow is extended or flexed as shown in **Figure 27.** The muscle moment in both **Figure 27a** and **27b** is a flexion moment but the arm is flexing in (*a*) and extending in (*b*). The power expended by the muscle in both cases is the product of the moment and the angular velocity of the forearm, but in case (*a*), the angular velocity is positive (in the same direction as the muscle moment), and in case (*b*), the angular velocity is negative (in the opposite direction of the muscle moment). The power in case (*a*) is positive and in case (**b**), it is negative. When examining the muscle activity, it is evident that in case (*a*), the muscle is doing concentric contraction, or positive work, and in case (*b*), the muscle is doing eccentric contraction (lengthening), or doing negative work.

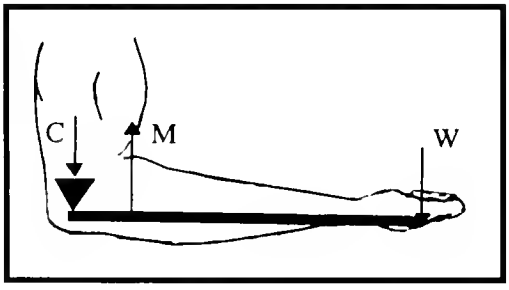

Figure 26.
Lever model of forearm.

Figure 28 shows the angle dorsiflexion and plantarflexion of the ankle joint during stance phase of gait where dorsiflexion is plotted positive, therein illustrating how the power can be used to understand the muscle activity. The ankle first plantarflexes as the GRF acts posterior to the ankle joint and then dorsiflexes as the center of mass passes over the foot and finally plantarflexes until toe-off. The muscle moment during stance phase is approximated by **Figure 29.** During the initial period of stance, the tibialis anterior is active as the foot is plantarflexing to foot-flat. During the rest of stance phase, the gastrocsoleous muscles are active controlling the center of mass of the body as it passes over the foot and then providing the power to push off the body and transfer the weight to the opposite foot. The power of the muscles is shown in **Figure 30.** The initial power is negative as the tibialis anterior muscles break the foot during foot-fall and the power is again negative as the ankle plantar flexors control the dorsiflexion of the foot. The final power is positive as the plantar flexors go into concentric contraction to power the body up and forward.

Overuse injuries are usually associated with positive power output by the muscles (i.e., when the muscles are being used to produce positive work on the body). Examples of this are pushing off while running, jumping, rising during squatting, and other concentric contractions of the muscles. Trauma injuries usually occur when the power of the muscles is negative or the muscles are trying to break an external moment that is being applied. A common example of this is when a runner hits a pothole in the road and the GRF occurs on the forefoot instead of the heel. The plantar flexors are forced to try to resist a suddenly applied dorsiflexion moment, usually of a high magnitude applied at a rapid rate, and the heel cord cannot

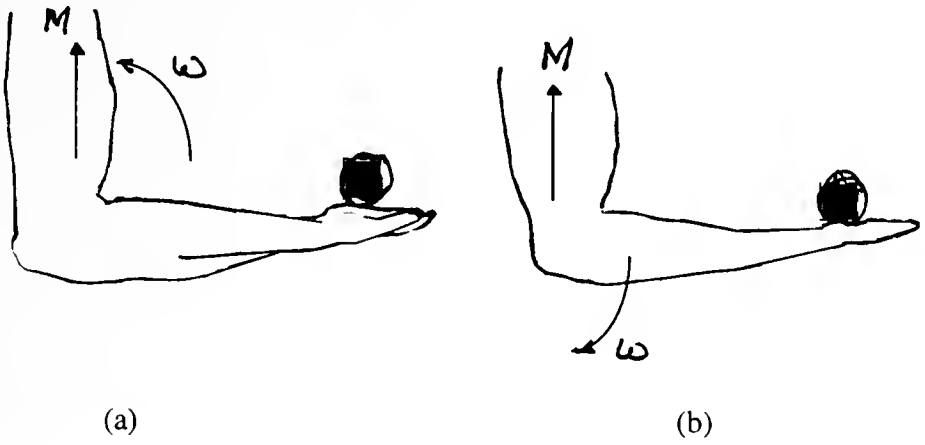

Figure 27.
Muscle power during elbow flexing or extending.

withstand that high strain rate, resulting in a partial tear or complete rupture.

APPLICATIONS OF MOTION ANALYSIS AND BIOMECHANICS

The applications of biodynamics to the medical field are increasing as research in this area is expanded and new equipment becomes available. The overall purpose of biodynamics is to provide a quantitative measurement of the function/dysfunction of the neuromuscular skeletal system. It is the responsibility of the biomechanist in cooperation with the clinician to provide information that cannot be obtained by other methods and, more importantly, to establish the clinical relevance of this information.

The most common application of motion analysis and biomechanics is gait analysis; for this reason, most clinical and research laboratories are called gait laboratories. However, to limit the application of this equipment and the analytical tools to gait analysis would be a failure to understand the full range of medical applications. One common application is the assessment of postural balance. Although balance is understood in a lay sense as the ability to maintain physical equilibrium, it is necessary to define it in a mathematical sense if biomechanical measurements are being made. When standing quietly, an individual is said to be in perfect balance when the center of gravity (c.g.) of the body is

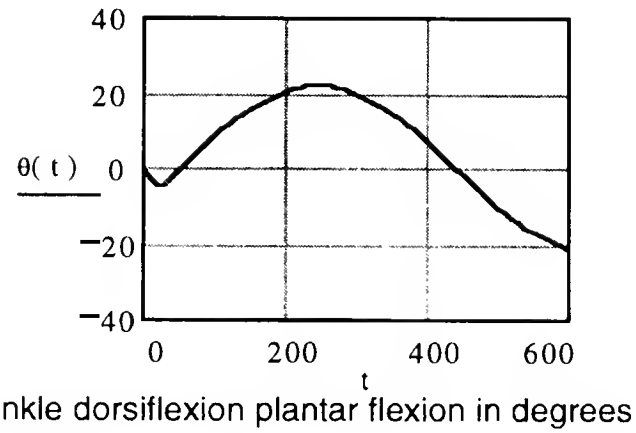

Figure 28.
Ankle dorsi/plantar flexion angular position.

directly over the center of pressure (COP) of the GRF. This is illustrated in **Figure 31.** The COP is measured with the force plate and can be accurately determined at any instant of time. The difficulty in obtaining rapid balance assessments is due to the inability to define the location of the center of mass in real time. An 11-segment model of the body may be made comprising the head, trunk, two upper arms, two forearms, pelvis, two thighs, and two lower legs. Data are available for the proportion of mass of each segment and the location of the center of mass of each segment for men and women of different sizes. However, accurate establish-

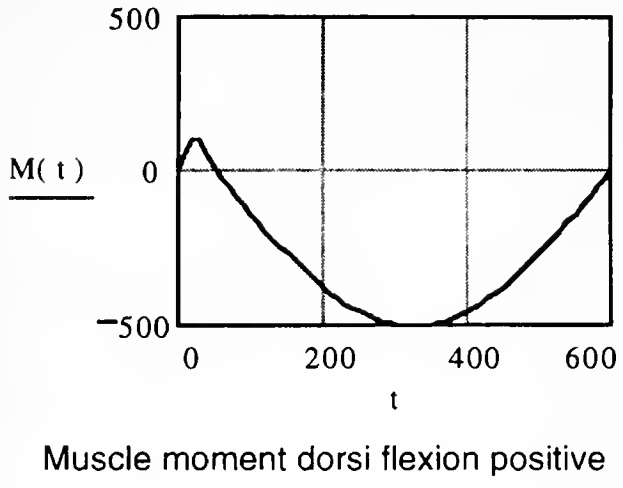

Figure 29.
Ankle muscle moment.

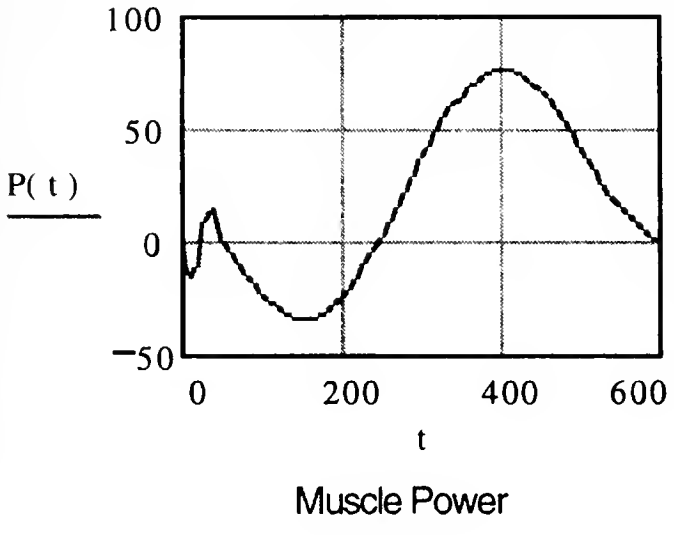

Figure 30.
Ankle muscle power.

ment of the position of each of these segments requires three markers per segment, totaling 33 markers.

The most common protocol to measure postural stability uses only a force plate. The COP is measured over a specific time frame, usually 20 seconds, and data are sampled at a specified rate, that is, 50 to 100 Hz. The resulting plot of the COP is called a stabilogram (see example in **Figure 32**).

The mean radius of the stabilogram can be computed and the velocity of the movement of the COP determined. These have been used as measures of

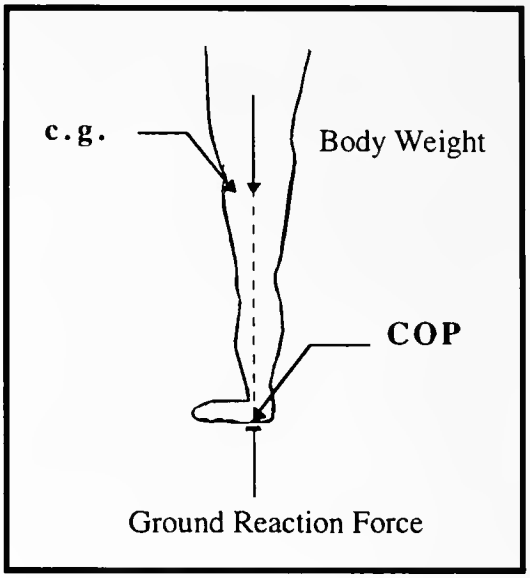

Figure 31.
Postural balance model.

postural stability. The COP must remain within the area of the base of support (the area underneath and between the feet), which decreases as the stance width decreases and increases as the feet are moved farther apart to a more stable position. Data have been collected for individuals standing with feet together, feet apart, feet tandem, or standing on one leg. In addition, eyes open and eyes closed data give the influence of vision on postural stability. Some laboratories have had the individual stand on soft mats to obtain information on proprioceptor influence on balance.

The major difficulty in using only the COP as a measure of balance is that the subject can move the position of the COP anteriorly, posteriorly, or laterally and still remain in a controlled balance position. Therefore, the data are dependent upon the individual trying to maintain balance and not deliberately perturbing the stabilogram. This is not the case when the difference between the c.g. and the COP is used as the measure of balance. Simpler models to determine the c.g. of the body are being introduced to obtain better measurement of postural balance and still maintain a workable laboratory protocol.

The ultimate goal of biodynamics is the development of predictive computer models for the neuromuscular skeletal system. This would allow the evaluation of an individual using the techniques of the indirect

dynamics model to determine joint motion and muscular activity. A computer model would then be created that could be driven using the muscle moments at the joints. The principal difficulty of models of this nature arises in the nonlinearity of the differential equations of motion. The solution of these equations becomes unstable as the solution progresses. Another problem is the over-determinacy of the number of muscle forces that act across a joint, that is, there are more flexors or extensors than are needed to flex or extend the joint. Using EMG, the temporal activity of these muscles may be determined but the reason for this over-determinacy is not fully understood. This problem is frequently approached using the mathematical methods of optimization.

In the simple model of the forearm shown in **Figure 33**, consider M_1 and M_2 as the forces in muscles 1 and 2 that flex the elbow. Either of these two flexors would be sufficient to flex the elbow or maintain equilibrium of the forearm when the weight **W** is applied. The equilibrium equation is

$$M_1a + M_2b = Wc \qquad [19]$$

Note that the model of the forearm shown in **Figure 26** lumped the forces of the two muscles together as one equivalent muscle. Equation 19 is the only relevant equilibrium equation to determine both of the muscle forces, so there is no more information available from traditional biomechanics methods. EMG data will show that both of these muscles are active while holding the weight in the hand. Note that this model has been simplified and antagonistic muscle activity is not even considered.

Optimization techniques search for cost functions that should assume a minimal value to optimize the system. The earliest cost function that was introduced to optimize joint function was the total muscle force active at any time, which is

$$M_T = M_1 + M_2 \qquad [20]$$

Therefore, Equation 19 would have to be satisfied while the total muscle force was minimized. This can be illustrated in **Figure 34**, where the total muscle force is plotted against the value of the two muscle forces (muscle space) and is shown as a shaded surface. Since the muscle forces must satisfy the equilibrium condition (Equation 19), the only solution values for the relative values of the muscle forces must lie on the line that represents that equation. Search along this line to find the point where the total muscle force is a minimum.

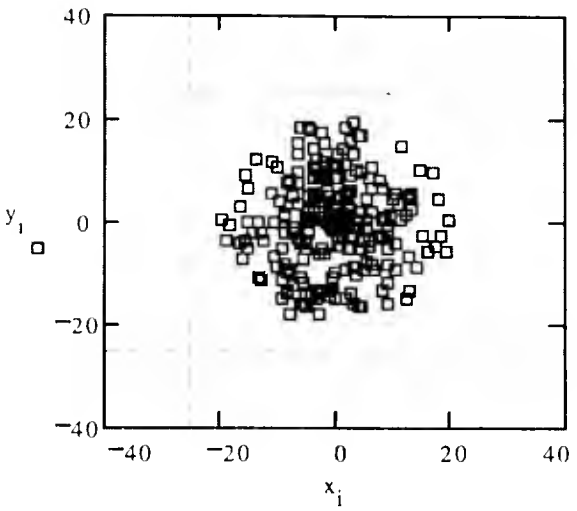

Figure 32.
Postural stabilogram.

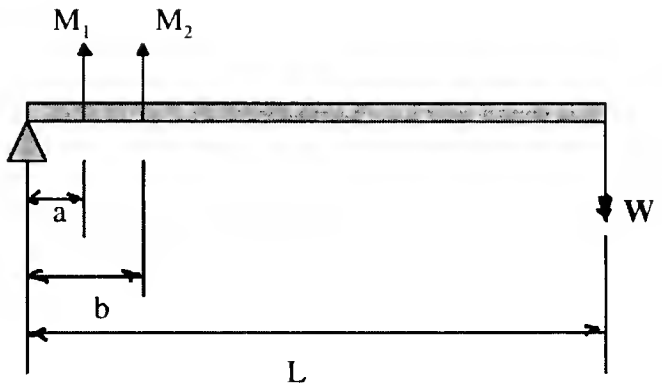

Figure 33.
Two muscle lever model of forearm.

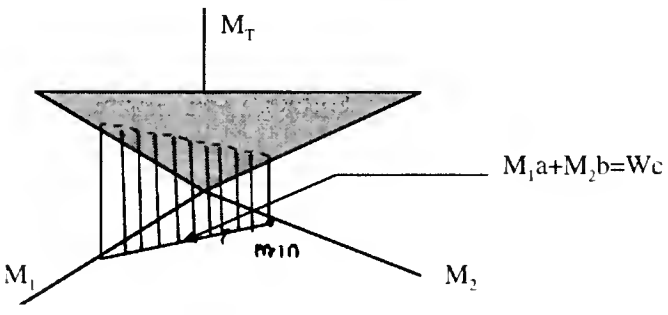

Figure 34.
Optimization of forearm muscles using a total muscle force cost function.

The optimum value of the muscle forces that minimize this cost function is such that M_1 would not be active and equilibrium would be maintained by use only of the muscle that had the largest moment arm or mechanical advantage. Experiments show that this is not the manner that the body uses to maintain equilibrium. Therefore, the cost function of the total muscle force is not the explanation of the over-employment of muscles during any given activity.

New cost functions have been investigated, such as the total muscle stress (muscle force divided by the cross-sectional area of the muscle), total muscle energy, and physiological fatigue of muscles. If an appropriate optimization model can be obtained, then the clinician can rehabilitate the patient in conjunction with the body's own attempt to minimize certain physiological costs.

Motion analysis and biomechanics are also currently being used to measure upper limb function, spinal curvature, and cervical spine movement. There are detailed studies to evaluate different orthopaedic surgeries, various instrumentation for total joint replacement, and to functionally evaluate different prosthetics and orthotics. Quantified functional assessment is also valuable in evaluating pharmaceutical drugs used for muscular skeletal problems.

REFERENCES

1. Holden JP, Stanhope SJ, Orsini JA. Skeletal motion estimates: effect of surface target techniques. Proceedings of the Second World Congress of Biomechanics, Amsterdam, The Netherlands; 1994;(2):372.
2. Goldstein H. Classical mechanics. Reading, MA: Wesley Publishing Company; 1950. p. 144–8.
3. Grood ES, Suntay WJ. A joint coordinate system for the clinical description of three-dimensional motions. J Biomech Eng Trans ASME 1983;105:136–44.

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SECTION THREE

Chapter One

The Value of Information Resulting from Instrumented Gait Analysis: The Physiatrist

by Andrew Gitter, MD and Robert McAnelly, MD

Dr. Gitter is an Associate Professor and Dr. McAnelly is an Assistant Professor, at the University of Texas Health Science Center at San Antonio Rehabilitation Medicine, San Antonio, Texas.

INTRODUCTION

Gait analysis has a long history and tradition, from the pioneering work of Muybridge and Inman continuing through contemporary times with the development of modern computer-based analysis systems capable of describing the kinematic, kinetic, and muscle activation patterns of gait in unusually rich detail. Over the past several decades, instrumented gait analysis has emerged as a powerful tool in the research setting. Through descriptive and experimental studies, gait analysis has advanced our understanding of normal gait, identified and quantified the biomechanical and motor control abnormalities of pathologic gait, and documented the usefulness of various therapeutic interventions. In contrast to the established role that quantitative gait analysis has achieved as a research tool, the clinical use of gait laboratories and gait analysis by physiatrists and other rehabilitation care providers is uncommon. With the exception of diagnostic and surgical planning purposes in children with spastic paralysis, which has been largely driven by orthopedic surgeons, instrumented quantitative gait analysis has not been systematically adopted for the evaluation of gait in other patient populations. In the rehabilitation literature, there are intriguing case reports of gait analysis improving patient care, and evidence that instrumented gait studies can aid in the diagnosis and determination of the pathomechanics of some gait abnormalities. However, there is not a substantive body of data that clearly identifies the groups of patients, or the gait abnormalities commonly managed by physiatrists in which instrumented studies are beneficial to overall care and function. Perhaps because of this, efforts by proponents to expand its role in the management of adults with disabling gait problems from a variety of neurological and musculoskeletal disorders has met with limited success. Moreover, and perhaps more troubling, there is little spontaneous interest or call for expanding the use of this technology by most physiatrists.

The current state of clinical gait analysis in the practice of medical rehabilitation raises several important and timely issues that the physiatrist needs to consider. These are best addressed by clearly separating the research role of gait analysis from its use as a clinical procedure. In light of the limited data and uncertainty over the role of gait analysis in the care of adults with impaired ambulation, this chapter will focus on the important conceptual and practical barriers that limit its clinical use by the physiatrist. The barriers and limitations listed in **Table 1** combine recommendations from a recent National Center for Medical Rehabilita-

Table 1.
Barriers, limitations, and unanswered questions concerning the use of clinical instrumented gait analysis by the physiatrist.

Lack of objective data that instrumented gait analysis improves patient function.

- Effect of gait analysis on diagnosis, clinical decision making, and treatment selection is unclear
- Lack of cost-effectiveness information

Limited information or guidelines for selecting and applying specific gait analysis techniques in evaluating and treating different gait abnormalities.

- Is standardization of gait analysis protocols for different disorders useful?
- Better definition of the patient populations and gait problems that are benefited by instrumented gait analysis.
- Does instrumented "motion" analysis improve the care of nonambulatory mobility problems or upper limb motor disability?

Limited treatment options for use in the management of adult gait disorders.

- Current physiatric interventions are empirically based and have low morbidity, lessening the need for instrumented gait analysis.
- Improved neuromuscular and musculoskeletal models of gait needed to allow prediction of compensatory strategies and treatment outcomes.

Limited understanding by clinicians of the data generated by instrumented gait analysis.

- Better training of residents and clinicians in the complexities of the kinematic, kinetic, and motor control features of gait
- Improved gait educational media
- Standardization of terminology to improve communication

tion Research sponsored workshop on the future of gait analysis (1) with those of other researchers (2–4) and personal observations. These barriers touch on multiple aspects of gait analysis: basic technology concerns over the ease and accuracy of data acquisition, uncertainty regarding the value of different gait measures in various disorders, and fundamental concerns over its clinical effectiveness.

An initial, useful perspective can be gained by reviewing the gait issues associated with the assessment of children with cerebral palsy (CP), the disorder in which gait analysis has achieved its greatest level of clinical acceptance. By characterizing the reasons for its relative success in *this* population, the limitations and problems that have prevented its use in the *adult* population become more clinically apparent.

The neuromuscular manifestations of CP are heterogeneous. A wide spectrum of clinical gait disorders is present in children with CP that ranges from unilateral spastic hemiplegic gait to the diplegic crouched gait pattern (5,6). Altered central nervous system motor control of gait is superimposed upon varying degrees of muscle or joint contracture and (mal)adaptive changes in skeletal growth and alignment. The result is a dynamically evolving gait pattern in a growing child caused by the complex interplay of abnormal muscle timing and force generation, secondary limitations in joint range of motion, and altered muscle force lever arms caused by skeletal adaptations of the lower limb joints. Treatment of these abnormalities involves the collaborative efforts of multiple health care providers and may include: 1) various surgical procedures done at appropriate times during the child's development, 2) the use of serial orthotic devices, 3) both invasive and noninvasive spasticity management, and 4) physical therapy. Successful optimization of a child's gait can lead to a lifetime of improved mobility, function, and quality of life, while inappropriate treatment may worsen disability.

In summary, the gait disorders experienced by children with CP are heterogeneous, complex, and involve invasive treatments, but can offer a lifetime of improved function. The use of gait analysis to characterize a child's walking pattern is intuitively rational and improves the understanding and inter-relationships between multiple complex factors unique to each child. Analysis allows the longitudinal tracking of the evolution of gait, and can assess treatment effects. There is some evidence that supports the use of surgical interventions in improving gait (7), demonstrates that gait analysis alters surgical decision-making (8), and improves clinical outcomes (9). Yet, despite the use of gait analysis in spastic paralysis for the past decade and strong advocacy supporting its use, considerable controversy still exists over its true clinical value as highlighted in the recent editorials by Gage (2) and Watts (10).

In many respects, CP gait encompasses a unique constellation of clinical characteristics that is more complex than the issues surrounding the management of gait in most adult rehabilitation settings. Hemiplegic gait following brain injury is likely the most common central nervous system gait disorder that the physiatrist must manage and serves as a useful model for understanding the role of instrumented gait analysis. Several major differences will be highlighted in this

Chapter. First, studies have shown that the altered gait motor control in hemiplegia usually falls into one of three patterns: premature activation of the plantarflexors, reduced activity in one muscle or in muscles groups, and co-contraction (11–13). As a result, the adult hemiplegic gait pattern is more stereotypic than the diversity of abnormalities seen in children with CP. The resulting biomechanical deficits that compromise walking primarily affect knee and ankle control. Successful treatment strategies using combinations of foot, ankle, and knee orthotic devices; upper limb assistive aids; neurolytic procedures; and functional training have long been accepted as the standard of care. The principles underlying the use of orthoses and neurolytic procedures are based on generally accepted but simplified biomechanics of moment generation at the ankle and knee. While there has been some objective verification of the effectiveness of current treatments (14,15), we do not know if contemporary clinical practice maximizes gait ability. Secondly, the bony adaptations and developmental changes seen in the maturing skeleton of the child with CP are not a clinical concern in the adult. Thirdly, the surgical interventions used in these children are only rarely used in adults. Finally, expectations for what is an acceptable gait, perhaps not always to the benefit of individuals, are lower in the adult, especially the elderly person with stroke. This combination of clinical features surrounding adult hemiplegic gait has lead to a general acceptance of a relatively nonaggressive, noninvasive treatment paradigm that has not changed substantially for many years and typically does not attempt to understand the gait disorder with the degree of detail that can be obtained with instrumented gait studies.

Issues That Must be Addressed Before Clinical Gait Analysis Will be Used by Physiatrists in the Treatment of Adult Gait Disorders

Lack of Objective Data that Instrumented Gait Analysis Improves Diagnosis and Treatment Outcomes Over Standard Visual Observation Techniques

The lack of convincing data that instrumented gait analysis is more effective in improving diagnosis or treatment outcomes than standard clinical visual observation techniques is one of the most important issues that needs to be addressed. Visual observation of gait is the clinical standard and trained observers typically believe they can recognize many gait deviations, correctly assess the cause of the deviation, and infer an appropriate treatment strategy (16). This is probably true in gait problems resulting from musculoskeletal or peripheral neurologic disorders isolated to a single joint or nerve, but the reliability of visual observation in the more complex gait disorder associated with central motor control abnormalities is probably poor. The limited number of studies on observational gait analysis supports this concern by consistently showing that the interobserver reliability of visually identified kinematic deviations is only fair to moderate (17–19). If reliability is only fair in simply identifying motion abnormalities, it is undoubtedly even less reliable in accurately predicting the timing or pattern of alterations in muscle activation and/or the joint kinetics that underlie the movement disturbances (20). The limited reliability of visual observation would argue, empirically and intuitively, for an expanded role of instrumented gait analysis. Before this can be accepted, two issues need to be resolved: 1) whether instrumented gait analysis is more reliable and reproducible than visual gait analysis and 2) whether the increased time, effort, and expense of instrumented gait analysis affect treatment decision-making and functional outcome.

The reliability and reproducibility of instrumented gait analysis has received only limited study. Basic intertest reproducibility of kinetic and kinematic measurements appears to be adequate for clinical purposes (21,22). Empirically and intuitively, it is believed that quantitative gait information would improve interexaminer reliability in the identification of gait abnormalities, but this has not been adequately studied. The reliability of expert interpretation of gait studies and subsequent treatment recommendations is unknown but has been questioned (10).

Little published data directly address the effect of gait analysis on changing treatment or altering functional outcomes. While limited data exist showing that gait analysis improves the management of children with CP (8,9), clinicians must use case reports and indirect evidence to determine the value of gait analysis in other gait disorders. Case reports have highlighted the value of gait analysis in selected individuals (23–25) but do not constitute sufficient evidence to justify its widespread, general use. Examples of indirect evidence supporting the clinical use of gait analysis come from studies that suggest quantitative gait examination can identify abnormal and possibly injurious joint force development (26), predict response to botulinum toxin use in spasticity control during walking (11), and aid in

the choice of the most appropriate therapeutic technique or orthotic intervention in stroke (20). The lack of a substantive body of evidence of its clinical value is surprising and somewhat worrisome given that it has been available as a clinical procedure for nearly a decade. This is especially problematic given the current financial pressures on health care systems to justify the use of any new or expensive intervention.

Limited Information and Guidelines for Selecting and Applying Specific Gait Analysis Techniques in the Evaluation and Treatment of Different Gait Abnormalities

Gait analysis encompasses a wide range of measurement technologies designed to capture and characterize the temporal, spatial, kinematic, kinetic, and muscle activation pattern of an individual's gait. Kinematic procedures measure the motion of the body and limb segments through space during representative walking strides. Markers are placed over predefined bony landmarks on the arms, trunk, pelvis, and legs. The markers are used with a variety of image-capture technologies to track the three-dimensional locations of individual body segments throughout a gait cycle. From this raw coordinate data, joint range of motion and angular velocities are calculated for clinical analysis.

Kinetic analysis is used to determine the net forces and torques (moment) exerted on the body as a result of the combined effects of the ground reaction force, inertia, and muscle contraction. Kinetic analysis requires the simultaneous (i.e., during the same gait cycle) collection of kinematic information and ground reaction forces. Ground reaction forces are collected as subjects walk over force plates embedded into the floor of the laboratory. The calculation of the forces and moments generated at each joint is based on inverse dynamics physics and simplified models of the musculoskeletal system.

Dynamic electromyography (EMG) is used to determine the timing of muscle activation and to crudely estimate the relative magnitude of muscle contraction. EMG data can be collected using surface electrodes or, when greater muscle specificity is needed, intramuscular wire electrodes. The EMG signal is amplified and transmitted via telemetry or cable to a central computer where it is synchronized with kinematic and kinetic data, thus allowing inference about the muscular sources of force and motion abnormalities.

Separating the cause(s) of an abnormal gait from adaptive and potentially beneficial compensatory strategies used by an individual is not necessarily straightforward. The multiplicity of the central and peripheral mechanisms associated with the control of gait leads to a degree of indeterminacy in understanding any particular gait pattern. This is made worse by the absence of good models of the neuromuscular control strategies adopted by persons with different disorders that affect walking. Thus, there is a tendency to collect the entire spectrum of gait information in order to maximize the likelihood of measuring the relevant and important discriminating kinematic, kinetic, and muscle-timing features of a particular gait pattern. From a practical standpoint, this adds to the cost, complexity, and time required for gait analysis, especially in pathologic gait situations where increased stride-to-stride variability, balance deficits, and/or cognitive limitations interfere with data collection. The relative importance of the various subcomponents of gait analysis—kinematic, kinetic, and EMG, either individually or in combinations to the diagnosis or treatment of different gait abnormalities—is unknown.

The situation is analogous to the electrodiagnostic evaluation that is performed to "rule out neuromuscular disease." In the absence of a more specific clinical question, testing tends to be extensive, poorly focused, time consuming, and often of unclear clinical utility. Instrumented gait analysis may achieve greater clinical acceptance and be more cost effective if analyses can be focused on answering specific clinical questions. As a hypothetical example, consider the person with genu recurvatum following a traumatic brain injury. Recurvatum may result from several motor control abnormalities (premature plantarflexor muscle activity, prolonged quadriceps activation) or as overcompensation for absent quadriceps activity. To distinguish between clinically relevant causes, a directed gait study might only require dynamic EMG recording from the quadriceps and plantarflexor muscles along with footswitch information to determine the timing of muscle activation relative to heelstrike and toeoff. Such a gait study would obviously not completely characterize the individual's gait or allow comment on other potentially treatable abnormalities, but would be simple, require little time, and would likely be more cost effective.

Limited Treatment Options for Use in the Management of Adult Gait Disorders

The interventions commonly recommended by physiatrists for treating gait disorders can be broadly classified into one of several approaches: 1) physical

therapy based task training to improve functional skills; 2) orthotic, prosthetic, and assistive devices to improve balance, alter biomechanical forces, or control joint positioning; and 3) spasticity management with systemic drugs or local neurolytic procedures.

Physical therapy is seldom if ever detrimental but the selection of specific modalities or techniques is empirical and not well-based on objective information. It is not known whether gait analysis can help in choosing the best treatment approach, predict response to treatment, or determine if maximal recovery has occurred. Lower limb orthotic devices have seen substantial evolutionary advances in materials and options over the past several decades but there has been little fundamental improvement in their effect on the abnormal biomechanical or neurophysiologic features of gait. Prescription is based on well-accepted biomechanical principles, which can be modified as needed and, when clinical uncertainty exists, can incorporate adjustable joints to allow empirical gait optimization. How will gait analysis improve the prescription of lower limb orthoses: better device selection, defining optimal joint position, identifying persons for whom orthoses are inappropriate? These are questions for which answers do not yet exist. Neurolytic procedures, while variably successful and difficult to titrate, can be safely performed, especially if limited to motor point or motor nerve blocks. The effects of blocks are generally limited to months, lessening the risk of any permanent unexpected adverse effects on gait. When uncertainty exists, is it easier and more efficient for patients and clinicians to perform temporary local anesthetic blocks that may give both diagnostic and therapeutic information than to perform an instrumented gait analysis? How much of the advantage of "instrumented gait analysis" comes from quantitative measurements as compared with simply the clinical evaluation by an experienced consultant/expert in gait?

For the adult with a major disability, gait dysfunction is usually only one aspect of the overall impaired functioning. The emphasis and effort placed on improving gait is more or less important depending on other coexisting cognitive, sensorimotor, and pyschosocial problems. When the clinical features of currently available rehabilitation interventions (low acute morbidity, limited risk of long-term adverse sequalae, empirical application, and variable impact on overall function) are combined with uncertainty about the effectiveness of gait analysis in improving treatment and outcome, the current standard of care based on simple clinical assessment and judgment appears clinically rational and appropriate. The risk to continuing this seemingly appropriate observational approach is the lost ability and opportunity to critique and measure our treatment effectiveness, to uncover their limitations, and to encourage us to develop better therapeutic strategies.

Justifying the use of instrumented gait analysis in the vast majority of patients will be difficult until there are better treatment options that either require greater selectivity in their application, place individuals at greater risk of injury or adverse effect, or are costly. Surgical procedures for tendon lengthening, release, or transfer have been recommended for selected adults but their use is sporadic and not generally available. A more systematic assessment of their utility in adults seems warranted. Recently, two new treatment modalities have become clinically available that may fit these criteria: botulinum toxin for treating local muscle spasticity and intrathecal baclofen pumps for use in persons with brain disorders. While the role for these treatments are currently being investigated, both potentially may be important advances in our ability to improve gait and mobility problems associated with increased muscle tone. These treatments are expensive and, in the case of baclofen pumps, invasive. The value of instrumented gait analysis in these settings is largely untested but may be useful in predicting therapeutic response (11) or as a tool for objectively documenting effectiveness, thus justifying the use or continued use of these interventions.

Limited Understanding by Clinicians of the Kinematic, Kinetic, and Electromyographic Data Generated by Instrumented Gait Analysis

Instrumented gait analysis can generate an overwhelming amount of data describing the complex temporal, spatial, and kinetic aspects of an individual gait pattern. Interpreting this information requires a detailed understanding of gait biomechanics, normal and abnormal patterns of motor control, and an ability to relate these features to the pathological motion that is observed during walking. Finally, integrating this understanding of the mechanisms underlying a gait abnormality into appropriate and useful clinical recommendations requires substantial experience. Gait analysis reports, even after being subjected to an interpretive summary by an expert, tend to be long and difficult to understand for the clinician without specific training or interest in

gait. This is not particularly unusual or unexpected for a highly specialized medical test, but does serve to further distance non-specialist clinicians from instrumented gait analysis and places them in the position of needing to act on information that they may not fully understand.

For physiatrists, this is unfortunate, since many of the patient populations that constitute the core of rehabilitation medicine have significant gait disability. Assuring expertise in the evaluation, diagnosis, treatment, and management of gait disorders is an important and integral part of maintaining control over this aspect of care. Achieving this level of expertise will require that contemporary concepts of gait be incorporated into training programs and continuing medical education programs, facilitated through clinical interactions with patients, and updated as new advances in gait therapeutics are developed. Contemporary general rehabilitation texts (27,28) that serve as a foundation for training and clinical care all include chapters on gait analysis, but do not adequately develop the necessary knowledge base that physiatrists need to adequately evaluate and understand the relationships between observational, kinematic, kinetic, and EMG aspects of gait. Instrumented gait analysis offers a unique and powerful tool for teaching these concepts and, in this context, is a vastly underutilized educational resource. Experts in gait need to develop more effective teaching methods and media for clinician education with one possible approach being the use of computer multimedia as demonstrated by Smith (29).

Raising the general level of awareness of gait biomechanics can improve clinical observational gait analysis skills and, at the same time, increase the awareness of the uncertainties inherent in current clinical approaches to gait evaluation and the need for more objective testing in selected individuals. An alternate approach to expanding the number of clinicians skilled in gait analysis is through the automation of analysis using artificial intelligence based gait diagnostic systems (30). This approach, while intriguing, will need to overcome the biases and difficulties other expert systems have had in achieving clinical acceptance and widespread use.

CONCLUSION

The research role of gait analysis in improving our understanding of the basic neurobiology and mechanics of gait and in assessing the value of new interventions seems assured. However, the role of instrumented gait analysis in the management of those individuals served by physiatrists faces an uncertain future. As the health care system finds itself under increasing pressure to financially justify the use of expensive diagnostic tools and treatment interventions, the lack of convincing data that expanding the use of gait analysis will improve patient function makes it difficult to argue forcefully for its use at this time. The path to increasing the role of clinical gait analysis lies in proving its value through additional case studies but more importantly through controlled studies demonstrating its effectiveness. Further development in gait laboratory technology to improve access, automation, ease of use, and cost is needed. Improving the education of clinicians in the quantitative pathomechanics of abnormal gait will not only improve traditional clinical care but will force clinicians into recognizing the ambiguities and limitations of visual observation, especially when costly or invasive treatments are involved.

REFERENCES

1. Recommendation on the future of gait analysis. Gait Analysis in Rehabilitation Medicine Workshop, National Center for Medical Rehabilitation Research, September 26, 1996. Arlington, VA.
2. Gage JR. The role of gait analysis in the treatment of cerebral palsy. J Pediatr Orthop 1994;14:701–2.
3. Czerniecki JC, Gitter AJ. Gait analysis in the amputee: has it helped the amputee or contributed to the development of improved prosthetic components. Gait Posture 1996;4:258–68.
4. Rechtien JJ, Gelblum JB, Haig AJ, Gitter AJ. Technology assessment: dynamic electromyography in gait and motion analysis. Muscle Nerve 1996;19:396–402.
5. Gage JR, DeLuca PA, Renshaw TS. Gait analysis: Principle and applications with emphasis on its use in cerebral palsy. J Bone Joint Surg 1995;77A:1607–23.
6. Sutherland DH, Davids JR. Common gait abnormalities of the knee in cerebral palsy. Clin Orthop 1993;:288:139–47.
7. Sutherland DH, Santi M, Abel MF. Treatment of stiff knee gait in cerebral palsy: a comparison by gait analysis of distal rectus femoris transfer versus proximal rectus release. J Pediatr Orthop 1990;10:433–41.
8. DeLuca PA, Ounpuu S, Rose SA, Sirkin R. Alteration in cerebral palsy surgical decision-making based on three-dimensional gait analysis (Abstract). Dev Med Child Neurol 1994;S70:7–8.
9. Lee EH, Goh JCH, Bose K. Value of gait analysis in the assessment of surgery in cerebral palsy. Arch Phys Med Rehabil 1992;73:642–46.
10. Watts HG. Gait laboratory analysis for preoperative decision making in spastic cerebral palsy: is all it's cracked up to be? J Pediatr Orthop 1994;14:703–4.

11. Hesse S, Krajnik J, Luecke D, Jahnke MT, Gregoric M, Mauritz KH. Ankle muscle activity before and after botulinum toxin therapy for lower limb extensor spasticity in chronic hemiplegic patients. Stroke 1996;27:455–60.
12. Knutsson E, Richards C. Different types of disturbed motor control in gait of hemiparetic patients. Brain 1979;102:405–30.
13. Shiavi R, Bugle HJ, Limbird T. Electromyographic gait assessment, Part 2: preliminary assessment of hemiparetic synergy patterns. J Rehabil Res Dev 1987;24(2):24–30.
14. Lehmann JF, Condon SA, Price R, deLateur BJ. Gait abnormalities in hemiplegia: their correction by ankle-foot orthoses. Arch Phys Med Rehabil 1987;68:763–71
15. Ounpuu S, Bell KJ, Davis RB, DeLuca PA. An evaluation of the posterior leaf spring orthosis using joint kinematics and kinetics. J Pediatr Orthop 1996;16:378–84.
16. Harris GF, Wertsch JJ. Procedures for gait analysis. Arch Phys Med Rehabil 1994;75:216–25.
17. Eastlack ME, Arvidson J, Snyder-Mackler L, Danoff JV, McGarvey CL. Interrater reliability of videotaped observational gait-analysis assessments. Phys Ther 1991;71:465–72.
18. Krebs DE, Edelstein JE, Fishman S. Reliability of observational kinematic gait analysis. Phys Ther 1985;65:1027–33.
19. Keenan AM, Bach TM. Video assessment of rearfoot movements during walking: a reliability study. Arch Phys Med Rehabil 1996;77:651–5.
20. Knutsson E. Can gait analysis improve gait training in stroke patients. Scand J Rehabil Med 1994;30(Suppl):73–80.
21. Kirkpatrick M, Wytch R, Cole G, Helms P. Is the objective assessment of cerebral palsy gait reproducible? J Pediatr Orthop 1994;14:705–8.
22. Kadaba MP, Ramakrishnan HK, Wooten ME, Gainey J, Gorton G, Cochran GVB. Repeatability of kinematic, kinetic and electromyographic data in normal adult gait. J Orthop Res 1989;8:849–60.
23. Kerrigan DC, Glenn MB. An illustration of clinical gait laboratory use to improve rehabilitation management. Am J Phys Med Rehabil 1994;73:421–7.
24. Abdulhadi HM, Kerrigan DC. Camptocormia: a biomechanical analysis. Am J Phys Med Rehabil 1996;75:310–3.
25. Forese L, Wootten M, Kadaba MP, McCann PD. Surgical management of equinovarus deformity in the adult with head injury. Orthop Rev 1993;22:1001–10.
26. Kerrigan DC, Deming LC, Holden MK. Knee recurvatum in gait: a study of associated knee biomechanics. Arch Phys Med Rehabil 1996;77:645–50.
27. Esquenazi A, Keenan MA. Gait analysis. In: DeLisa JA, editor. Rehabilitation medicine: principles and practice. Philadelphia: Lippincott Co.;1993. p. 122–30.
28. Pease WS, Quesade PM. Kinematics and kinetics of gait. In: Braddom RL, editor. Physical medicine and rehabilitation. Philadelphia: W.B. Saunders Co.;1996. p.83–103.
29. Smith PJ, Simon SR. Development of a gait interpretation, instruction, and report generation system. Rehabil R D Prog Rep 1996;33:46–7.
30. Wientraub MA, Bylander T, Simon SR. QUAWDS: a composite diagnostic system for gait analysis. Comput Methods Programs Biomed 1990;32:91–106.

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SECTION THREE

Chapter Two

An Overview of the Value of Information Resulting from Instrumented Gait Analysis for the Physical Therapist

by Sue Ann Sisto, PhD, PT

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INTRODUCTION

Instrumented gait analysis is frequently used today for clinical applications. Its role has expanded beyond the area of gait analysis to aid in clinical decision making for rehabilitation, surgery, adaptive devices, ergonomics, and athletics. Physical therapists are well trained in the area of gait and movement analysis and are, by definition, movement scientists. The evaluation of gait, locomotion, and balance includes a series of tests described in the *Guide to Physical Therapy Practice* (1). Entry-level didactic requirements include normal gait mechanics and pathological gait for nearly all disabilities. Those in physical therapy training primarily learn observational techniques for clinical gait analysis. In addition, some academic programs provide an introduction to various forms of instrumented gait analysis and their potential to augment the physical therapist's knowledge of gait. Instrumented gait analysis involves information about temporal and linear parameters during gait as well as joint angles, ground reaction forces, and muscle firing patterns.

The purpose of this chapter is to review the scope of quantitative gait analysis (QGA) as it pertains to the practice of physical therapy. A brief background about quantitative gait analysis is followed by a description of how physical therapists evaluate gait. An overview of gait analysis technology and goals of the information obtained from QGA is discussed. Groups likely to benefit from this information are presented. Limitations of gait lab data and suggestions for improvement are reviewed.

Background

Over the past 10 years, the clinical application of gait analysis has grown rapidly in the United States, as a result of several significant changes. The widespread development of user-friendly software allows the clinician to use instrumented gait analysis more easily. Hardware has also been refined to allow faster data acquisition and computing. These two advances have permitted the clinician to improve his/her understanding and interpretation of information provided by movement analysis technology.

The evaluation of children with spastic paralysis constitutes the onset of movement analysis for the clinical arena and continues to be a large focus of clinical gait analysis. Orthopedic surgeons frequently needed objective information about the success of their surgical interventions, compared to a pre-surgical assessment (2). Other clinical applications of instrumented gait analysis include the assessment of persons with stroke, Parkinson's disease and multiple sclerosis, spinal and orthopedic injuries or disease, and amputations.

Clinical use of the gait data takes the form of an evaluation, ideally before some intervention, followed by a summary of the biomechanical and neuromuscular influences of gait after an intervention. The intervention may be surgery, orthotic or prosthetic applications, and pharmacological or physical therapeutic treatments. The physical therapist can then interpret the clinical significance of the change in gait impairment as a result of such intervention.

METHODS

How Physical Therapists Evaluate Gait

Observational gait analysis (OGA) is defined as the visual inspection of walking. The identification and grading of gait deviations depends on the observer's experience and individual bias (3). The physical therapist observer is trained to see a range of gait events. The gross abnormalities are the most readily observed. Frequently, these are made more pronounced if the therapist asks the subject to walk at a higher velocity or removes some physical or mechanical assistance. The disadvantage of OGA over QGA is the tendency to focus the eye on the gross gait deviations while overlooking more subtle ones.

OGA is a preferable form of gait assessment when considering the ease, time efficiency, and low cost; however, several questions have been raised about its limitations (4–8). Studies evaluating the reliability and validity of OGA unanimously point to only a moderate level of reliability for intra- and interrater assessments (9). However, due to certain design differences in these studies, it is likely too early to eliminate OGA as an important clinical evaluation tool.

Systematic methods of gait analysis have been described by Perry (10) to establish standardization procedures within the field. Perry describes three steps to carry out systematic OGA: 1) organization and classification of the essential gait events, 2) anatomic sequence of observation to sort the multiple events at different joints, and 3) data interpretation for total limb function and for gait cycle differentiation.

Often a physical therapist will start the observation at the foot and assess distally to proximally as the foot hits the ground. Interpretation includes the influence of neuromusculoskeletal and/or behavioral factors that may produce a particular gait pattern, such as spasticity, pain, contracture, or lack of motivation. When OGA provides insufficient information about the etiology of gait deviations, instrumented gait analysis may be warranted.

Gait Analysis Terminology

There are several terms and definitions commonly used when studying gait analysis: *Gait kinematics* refers to the branch of mechanics that deals with joint angular changes over the gait cycle. Kinematics is evaluated by using external markers that can be observed by cameras. *Gait kinetics* is defined as the forces, moments, and powers that change over the gait cycle. These measurements are captured by the use of force plates embedded in a walkway. *Dynamic electromyography* (EMG) refers to the evaluation of muscle activity throughout the gait cycle. This is accomplished through the use of either surface or needle electrodes.

Gait Measurement Technology

Two-Dimensional versus Three-Dimensional Gait Analysis

Most of the current gait analysis systems today convert the two-dimensional (2-D) data from several cameras into three-dimensional (3-D) data to determine joint centers. This method allows for the measurement of gait when there is out-of-plane movement. Two-dimensional gait analysis is attractive to the clinician because there are fewer markers and cameras needed for data acquisition and processing time is significantly reduced.

When 2-D motion data reduction methods are employed for gait analysis, the assumption must be that the motion is planar. For example, the knee motion for the unimpaired person could be considered to move in a single axis flexion/extension plane. However, if genu valgus, or varus, or femoral, or tibial torsion is present, there is a distortion of the data because the joint plane is no longer parallel to the viewing plane.

Davis et al. (11) evaluated the differences between 2-D and 3-D gait analysis, determining that joint angles for the hip and knee were the most consistent, since these joints represent the smallest out-of-plane movement during normal gait. The ankle, however, had the greatest sensitivity between 2-D and 3-D gait analysis. This is not surprising since the ankle is oriented out of the sagittal plane externally by about 7–10°. Much literature on normal gait over the years must be considered with caution because the conclusions drawn were based on 2-D gait analysis. Davis and colleagues also extend this caution to the kinetic analysis.

Kinematics

The evolution of gait analysis technology has enabled the physical therapist to utilize the resultant data to aid the interpretation of locomotor performance. Since expensive motion analysis equipment is not likely to be present in physical therapy clinics, therapists use other methods of motion analysis, such as videotaping. This method creates a permanent record of gait performance that serves to document progress in therapy but still relies on the therapist's ability to observe gait in a reliable way.

Instrumented gait analysis systems, though not present in most physical therapy clinics, constitute the most prevalent method of motion measurement available to physical therapists. While electrogoniometers and accelerometers require less instrumentation, these devices are more often used in academic or research settings, because they are less user-friendly for the clinician. Clinical gait laboratories are becoming more widely present in hospital settings, and physical therapists can benefit from such quantitative data. For example, physical therapists are able to assess passive and active range of motion (ROM). It may be difficult, however, for the therapist to determine how this available ROM is incorporated into gait. A subject may present with a simple knee-flexion contracture; however, QGA provides the complete knee trajectory throughout the gait cycle to determine whether the subject is using 100 percent of the available ROM. The physical therapist can also determine to what degree the limitation in knee motion has affected hip and ankle motion at simultaneous time intervals. **Figure 1** is an example of the unimpaired and pathological knee flexion/extension trajectory: the right limb demonstrates excessive knee flexion at heel strike (0 percent) and peak knee flexion during the swing phase (80 percent). This pattern is indicative of insufficient quadriceps control or hyperactivity of the hamstrings at heel strike and insufficient dorsiflexion requiring excessive knee flexion for toe clearance during swing. This can be corroborated by examining the dynamic EMG at these intervals.

Ground Reaction Measurement

Force platforms embedded in a walkway provide information about the center of pressure or the point of application of the ground reaction force vector. The ground reaction force is the sum of all the forces of the body segments while the foot is in contact with the ground. Many gait labs are now equipped with foot pressure systems in addition to force platforms. This technology allows for the determination of how the load is distributed on the plantar surface of the foot. This information can be useful for individuals with orthopedic pathologies from foot/ankle fractures or those relating to disease processes such as diabetic neuropathy. **Figure 2** is an example of the normal ground reaction force during gait.

Another form of kinetic analysis involves the evaluation of joint moments and powers. Joint moments are the forces produced by muscles and ligaments acting at a distance from the joint center. Joint power is the net rate of generating (concentric contraction) or absorbing (eccentric contraction) energy by all muscles crossing a joint, and is the product of the joint moment and its angular velocity. If the calculation of joint moments is required, it is necessary to have a system that can have both kinetics and kinematics. Several investigators have used the examination of external joint moments to predict muscle forces (12).

Temporal and Distance Parameters

Step time and length, stride time and length, and walking velocity and cadence all fall under the definition of temporal and distance parameters of gait. Because physical therapists spend a great deal of time walking close to subjects to either provide verbal or physical assistance, they have a keen awareness of length and timing parameters, particularly when they are asymmetrical. There is a variety of easy methods for determining timing parameters when force platforms are not available. These include the use of a stop watch to determine walking velocity (distance over time), counting the number of steps per minute (cadence) over a known distance, chalking the soles of shoes or walking on carbon impregnated paper to determine step and stride lengths, or heel switches to determine step and stride time intervals. The greatest challenge to using heel switches is their placement when the subject does not achieve heel strike during gait. In this case, the distance parameters are distorted because the heel does not strike the ground first, thereby lengthening the step length parameter by a factor proportional to the length of the hindfoot.

However, these methods pose clinical challenges; namely, the clinician must have the time to obtain these measurements. A laboratory staff experienced in data collection and processing can provide the easiest method to obtain QGA results. The physical therapist can then be instrumental in the interpretation of

parameters such as asymmetry in joint trajectories and their etiologies. The therapist also receives feedback on how gait parameters change as a result of rehabilitation techniques. For example, although the therapist may be keenly aware of the impairments limiting asymmetry in the temporal and linear components of gait, it may not be obvious as to how much symmetry should be sought during treatment and what rate of treatment progression achieves the most efficient gait. Repeated measurements of the temporal and distance parameters from QGA may relate to a particular progression of gait training and may provide insight into optimal clinical practice guidelines. These studies are sorely needed to validate many interventions in the field of rehabilitation.

Dynamic Electromyography (EMG)

Voltage potentials detected by surface or wire electrodes provide information about the timing and intensity of the muscle contraction. Physical therapists know when muscle groups responsible for locomotion should be active, but it is most difficult to observe anything but gross muscle activity during gait; therefore, dynamic EMG can provide useful information unavailable by observation. Timing of muscle activity is the most frequently used parameter obtained from dynamic EMG. Weakness or spasticity of one or two joint muscles produces visible gait deviations, but it is difficult to observe whether the activity is present at the appropriate times. Neurological subjects who exhibit spasticity walk with a stiff-legged gait, and QGA provides information about which muscles are spastic and when they are misfiring (13).

The intensity of muscle contraction obtained from dynamic EMG is more controversial, because it must be normalized for some maximal effort, such as a maximal voluntary contraction. Instrumentation requires certain dynamic EMG systems to have gain settings that may alter the interpretation of the magnitude of the EMG signal. If the subject demonstrates spasticity, the maximal contraction is an abnormal response to voluntary contraction, making relative submaximal comparisons of muscle activity difficult.

Goals of Quantitative Gait Analysis

Comparisons to the Disabled Populations

Probably the most common use of instrumented gait analysis is the comparison of gait data from the disabled population to that of the nondisabled. An occasion where this normative external standard may be most useful is when there is an expectation for normal performance (2). The limitations of such comparisons include issues of age, gender, walking velocity, and anthropomorphic differences between the two groups. For example, comparisons of data from subjects who walk at different velocities from the normative standards have limited validity. Walking velocity is often drastically reduced in disability, and the normative data are most often collected at self-selected and faster-paced speeds. The effect of speed may explain a large portion of the group differences (14). We also know that men walk differently from women of similar height and

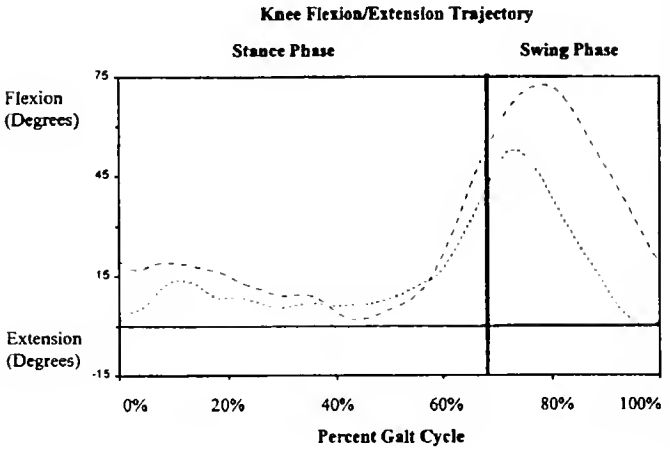

Figure 1.
Knee flexion and extension trajectory of a neurological subject during level walking at self-selected walking speeds. Pathological (.....), Normal (------).

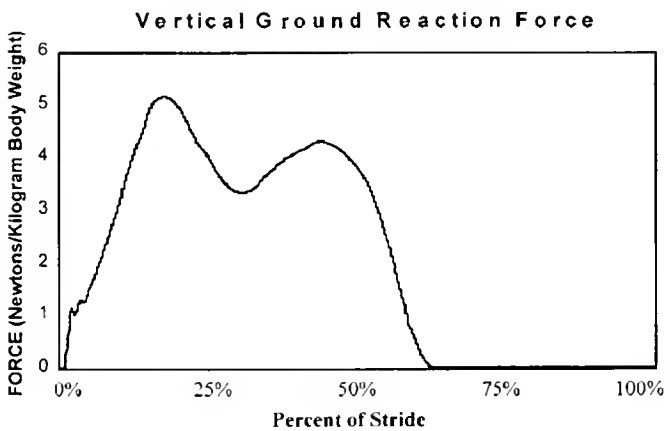

Figure 2.
Force output over a stride length (heel strike to heel strike on the same side) of a 45.36 kg female striking the force plate with the left limb while walking at a self-selected walking speed. Forces have been normalized by dividing by the body weight. Time base is normalized to 100% with toe-off at 60%.

weight and that age and maturation affect walking patterns (15). Therefore, caution must be used when making comparisons of disabled walking patterns to normative databases.

Identification of Disability Levels

As our healthcare environment is more often requiring outcomes that are quantitative, gait analysis can be useful in identifying levels of disability or functional loss. While therapists use measures of impairment to plan and implement treatment (16), it is important to relate these impairment measures to disability levels. QGA is well suited to provide information about the impact of impairments during a functional movement. This is an important service, as physical therapists make very well defined impairment measures in a "static" position but have to interpret how these impairments relate to functional problems in walking.

Several studies have begun to address the identification of disability levels using QGA. Knutsson (17) used dynamic EMG to classify subjects with hemiplegia through the assessment of spasticity patterns. Delitto et al. (18) classified those with low back pain, and Richards et al. (19) classified the recovery stage of persons who had suffered a stroke, based on their walking velocity. These studies effectively relate the gait impairments to the disability outcome and make comparisons within a very homogeneous group rather than comparing the data with those of a normative database.

Efficacy of Interventions

Currently, there is a great demand to shorten the length of hospital stay and a concomitant demand to increase function: these demands require practitioners to seek the most effective treatment interventions. Physical therapists have many treatment paradigms at their disposal, but there is little information available regarding the intensity and duration of treatment. Gait analysis can be a valuable tool to determine the benefit of mobility treatments and their transference to the functional task of walking. QGA comparisons made before and after treatment can aid in the determination of the intensity and duration of treatment, and when the effect of a treatment has reached a plateau. For example, using QGA, conclusions may be drawn about which surgical or physical rehabilitation techniques for the person with anterior cruciate ligament deficiency (ACLD) are beneficial. Sinkjaer and colleagues studied muscle coordination of ACLD subjects and classified those who were good versus those who were poor compensators during walking (20). These differences have important clinical treatment implications but would not have been identified with a less sophisticated form of analysis.

Performance Enhancement

Physical therapists often render treatment to improve normal performance, as is the case with athletes. Williams et al. (21) conducted a study using gait analysis coupled with oxygen consumption measurements to compare recreational and elite runners. These authors reported that gait analysis proved to be more useful than the assessment of running efficiency. Therefore, gait analysis may be a useful tool to predict the incidence of injury in athletes. This predictive information is extremely desirable to third party payers and helps to justify prophylactic physical therapy treatments and the promotion of wellness in the athletic population.

Mechanisms of Gait Deviations

A major challenge for physical therapists in the evaluation of gait is the determination of which deviations are primary and which compensatory. Observational gait analysis provides one answer to the trained observer; however, these judgments need to be corroborated by quantitative measures. For example, the contribution of muscles to measured motion versus passive movement during the gait cycle has been evaluated. A study by Sutherland et al. (22) indicated that the plantar flexors are accelerators, but only in 40–50 percent of the gait cycle and not at terminal stance as was originally thought. From a therapeutic point of view, it is quite possible to observe sufficient plantar flexion that may be occurring later in the stance phase (60 percent and beyond). This observation may lead the physical therapist to decide that there is sufficient plantar flexion power, not realizing that the plantar flexion is occurring late in the gait cycle and is likely a result of the passive extension at pre-swing due to a rapid unloading of the limb. Once the source of a gait deviation is determined to be primary, a physical therapist is better able to tackle the deficiency more directly.

Groups Potentially to Benefit

Anterior Cruciate Ligament Injury

The literature suggests that gait analysis might be helpful to physical therapists in determining whether an

individual is a candidate for a knee orthosis, rehabilitation, or surgery. Berchuck et al. (23) describe the gait pattern of persons with ACLD, which condition resulted in drastic changes in knee joint kinetics during level walking in the absence of serious kinematic differences. These researchers determined that a "quadriceps avoidance" pattern was evident. Kadaba et al. (24), on the other hand, describe the same phenomenon as increased-stance knee flexion that was balanced by quadriceps muscle activity. This activity was described as a positive adaptation due to the absence of the ACL. However, it becomes even more crucial when an anterior cruciate reconstruction fails over time. If physical therapists could receive feedback from kinetic analysis of the knee joint forces during weight-bearing exercises, alterations could be made in the treatment strategy to avoid overstretching a reconstructed ACL, and they could educate the patient as to proper functional activities to avoid reinjury.

Geriatric

Understanding the mobility problems in the elderly can point to rehabilitation strategies. Patla et al. (25) describe methods for classification and characterization of mobility performance, such as walking over a variety of terrains. Gait analysis can be useful in determining whether there are problems with the locomotor apparatus. The aging process is multifactorial; therefore, sensory and visual systems must be screened as well. Matching subjective reports with quantitative gait results may aid the physical therapist in determining which factors are the most limiting. For example, if pain is perceived during the stance phase and is correlated with reduction in force output of the quadriceps muscles or reduced knee moments and powers, the physical therapist can be guided by this information as to which impairments to address first in the rehabilitation program.

Pediatrics

Spastic paralysis often requires surgical intervention for the lengthening of spastic muscles and their tendons in an attempt to reduce muscle contractures and abnormal skeletal alignment. Without the knowledge of abnormalities of muscle function, floor reaction forces, and 3-D movements, functional outcomes had been disappointing. With the advent of clinical gait analysis, the American Academy for Cerebral Palsy and Developmental Medicine was the first to apply the results in the care of these persons (26).

Stroke and Head Injury

Central nervous system disorders can produce a mixture of spasticity, impaired motor control, and primitive reflexes, all of which can result in contracture. It is difficult to discern inappropriate muscle action during gait, because of the mass firing patterns and the compensatory efforts to control the limbs. Herein lies an extremely powerful benefit to the output from QGA for the physical therapist. The major muscle groups firing for extended periods are usually obvious to the skilled gait observer, but the relative contributions of muscles crossing the pelvis, hip, knee, and ankle are harder to determine. Dynamic EMG can clarify this relationship and guide the therapist in emphasizing manual interventions and gait training to reduce spasticity.

Gait EMG of spastic muscles has also helped in the clinical decision making for pharmaceutical interventions (27) such as Botulinum Toxin A (BTX). BTX paralyzes muscles firing at inappropriate times during gait and allows for strengthening of the antagonist muscle group and gait training without the influence of spasticity on the biomechanics of gait. The equinovarus deformity is a good example of the difficulty in differentiating the primary offending muscle during gait. The cause may be premature activity or contracture of the soleus, while tibial muscles display normal function. A blocking agent, such as BTX, used in a sequential fashion can help tease out the cause. The common assumption is that the tibialis posterior is the cause, but Wills et al. (28) discovered that this muscle was the primary cause in only 25 percent of the cases, while the tibialis anterior was the more frequent cause (45 percent) in 50 children with spastic paralysis. The physical therapist can use this diagnostic information to improve the balance of the firing patterns between the agonist and antagonist through neuromuscular reeducation, biofeedback, and/or strengthening.

Fatigue (Post-polio and Chronic Fatigue Syndromes)

Persons who have functioned well for 20–30 years since polio, are now experiencing disabling symptoms that are most often attributed to overuse, but the mechanisms are unclear (29). These individuals spontaneously substitute alternate muscles for the weak or absent musculature. Persons who have post-polio syndrome are primarily hampered by fatigue; therefore, an understanding of the intensity and firing pattern of the lower limb muscles is more appropriate than joint angle profiles. Typically, the firing patterns of muscle activity in these people are prolonged in order to avoid

instability during gait. The physical therapist who is aware of which muscles are particularly susceptible, will educate the subject about paced activity and provide gait-training techniques that incorporate as many alternate strategies as possible. Only one study has been conducted in the area of gait abnormalities in Chronic Fatigue Syndrome (30). This study identified gait abnormalities compared with sedentary healthy controls. However, the authors were not able to identify the causal factors for this apparent difference but hypothesized them to be due to balance problems, muscle weakness, or central nervous system dysfunction. Gait analysis has begun to characterize movement dysfunction in this population, but further work is needed to compare these deficits over prolonged exercise where the influence of fatigue may be more pronounced.

Prosthetics

Several studies have evaluated the effectiveness of prosthetic feet (31,32), many of which are designed to conserve energy. A study by Gitter and colleagues (32) indicated an improvement in energy efficiency of approximately 30–40 percent with the energy-storing feet over the standard SACH foot for persons with transtibial amputation walking at normal speed. However, the authors found that despite this greater mechanical performance, there were no significant differences in the knee and hip powers for the two feet. Typically, physical therapists and prosthetists will recommend the latest prosthetic components to enhance prosthetic gait, but may be unable to determine different energy requirements for walking with each component. Since physical therapists have an opportunity to assess energy requirements from subjective reports during prosthetic gait training, corroboration for this perceived decreased energy requirement from the use of sophisticated prosthetic components might be obtained from QGA. Torburn et al. (33) indicated that while the kinematic and kinetic patterns implied reduced energy expenditure during gait, oxygen consumption increased due to the increased intensity and duration of muscle firing. Dynamic EMG coupled with kinematic and kinetic gait analysis allows for a teasing out of discrete components incorporated into gait. A physical therapist is equipped to utilize this information to guide the treatment program and provide input for optimal prosthetic equipment.

Orthotics

Physical therapists commonly evaluate the appropriateness or effectiveness of an orthosis, particularly when an individual is in the acute recovery phase of rehabilitation. QGA, with and without the orthosis, reveals the degree to which an orthosis may help through the evaluation of temporal and linear parameters and kinematic trajectories. The physical therapist must include in the interpretation the physical constraints of the orthosis in evaluating the kinematic trajectories. For example, an ankle-foot orthosis may prohibit plantar flexion at heel strike but provide sufficient clearance during swing phase. The ankle trajectories will appear abnormal, due to this physical constraint. The decision about orthosis effectiveness might be better based on ambulation speed or temporal symmetry than on ankle trajectory that would appear abnormal from the QGA results. Therefore, while QGA provides information about gait performance, the interpretation of these findings is critical. The physical therapist can offer a therapeutic perspective as to prioritization of the gait parameters to which the subject is capable of adjusting during rehabilitation.

Limitations of Gait Lab Data

Improvement is needed in the quality of the data collected in gait laboratories. For example, if there is excessive soft tissue movement under the reflective markers, information about joint angle profiles may contain excessive variability that is not part of the movement pattern. This error of measurement is magnified when mathematical derivations are made to calculate velocity and acceleration. Further work is needed to improve the quality control efforts in practicing laboratories. Greater efforts must be made to standardize marker placement and terminology across gait laboratories to communicate effectively between clinical groups. There must be a better understanding of the underlying gait models utilized by commercial software, so that the interpretation of gait data considers the proper assumptions made by gait models.

Finally, normalization procedures need to be standardized to optimize across subject comparisons, in addition to the need for normative databases for homogeneous disabled groups. A common clinical use of gait analysis is to compare the performance of an individual with normative data to describe how the disabled gait differs from the "normal." We know that

certain anthropomorphic factors, such as weight and leg length, can vary the walking performance in nondisabled individuals. However, varying anthropomorphic features are not available in existing databases; if they were, the gait pattern of a subject of a specific stature could then be compared to a larger sample of persons with similar physical characteristics.

More definitive research that determines the utility of gait analysis in the clinical setting is needed. In most cases, OGA suits the clinician well for gross gait deviations. Even 2-D gait analysis may give sufficient information about gait performance with less complexity than is needed for 3-D analysis. Nevertheless, the system of analysis depends on the kind of clinical judgment the user intends to base on the data. If the physical therapist requires discrete unobservable information about walking performance, QGA will be needed. Currently, this situation would require a referral to a center where this level of analysis is performed, since it is impractical for most clinical settings. However, the more practitioners define a need for QGA, the more gait laboratories will be driven to provide information useful to the clinician in a timely manner. Clinicians seek to be successful with the patients under their care; therefore, whatever information aids in the planning of rehabilitation strategies will prove beneficial to them.

CONCLUSIONS

As all diagnostic and treatment methods will be under scrutiny during the medical economic readjustments currently underway, gait laboratories will have similar obligations. Clinical gait analysis will be obliged to relate objective findings to functional measures and outcomes. More importantly, in order for gait laboratories to survive in this arena, they will have to address how treatment progression can be guided by gait analysis. Physical therapists routinely address functional outcomes and are well equipped to assess the effectiveness of guided treatment as a result of quantitative findings about gait and movement analysis.

REFERENCES

1. American Physical Therapy Association's task force on practice parameters. What types of tests and measures do physical therapists use? Phys Ther 1997;77:1198–997.
2. Nene AV, Evans GA, Patrick JH. Simultaneous multiple operations for spastic diplegia: outcome and functional assessment of walking in 18 patients. 1993;75B(3):488–94.
3. Krebs DE, Edelstein JE, Fishman S. Reliability of observational kinematic gait analysis. Phys Ther 1985;65:1027–33.
4. Saleh M, Murdoch G. In defense of gait analysis observation and measurement in gait assessment. J Bone Joint Surg 1985;67B:237–41.
5. Miyazaki S, Kubota T. Quantification of gait abnormalities on the basis of continuous foot-force measurement: correlation between quantitative indices and visual rating. Med Biol Eng Comput 1984;22:70–6.
6. deBruin H, Russel DJ, Latter JE. Angle-angle diagrams in monitoring and quantification of gait patterns for children with cerebral palsy. Am J Phys Med 1982;61:176–92.
7. Patla AE, Clous SD. Visual assessment of human gait: reliability and validity. Rehabil Res 1998;1:87–96.
8. Eastlack ME, Arvidson J, Snyder-Mackler L, Danoff JV, McGarvey CL. Interrater reliability of videotaped observational gait-analysis assessments. Phys Ther 1991;71:465–72.
9. Malouin F. Observational gait analysis. In: Craik RL, Oatis CA, editors. Gait analysis: theory and application. St. Louis: Mosby-Year Book, Inc.; 1995. p. 112–24.
10. Perry J. Gait analysis systems. In: Gait analysis: normal and pathological function. NY: McGraw-Hill, Inc.; 1992. p. 351–4.
11. Davis RB, Ounpuu S, Tyburski DJ, DeLuca PA. A comparison of two-dimensional and three-dimensional techniques for the determination of joint rotation angles. Proceedings of the International Symposium on 3-D Analysis of Human Movement; 1991. p. 67–70.
12. Boccardi S, Pedotti A, Rodano R, Santambrogio GC. Evaluation of muscular moments at the lower limb joints by an on-line processing of kinematic data and ground reaction. J Biomech 1981;14:35–45.
13. Kerrigan DC, Gronley J, Perry J. Stiff-legged gait in spastic paresis: a study of quadriceps and hamstrings muscle activity. Am J Phys Med Rehabil 1991;70:294–300.
14. Murray MP, Drought AB, Kory RC. Walking patterns of normal men. J Bone Joint Surg 1964;46(A):335–60.
15. Finley RR, Cody KA, Finizie RV. Locomotion patterns in elderly women. Arch Phys Med 1969;50:140–6.
16. Dekker J, van Baar ME, Curfs EC, Kerssens JJ. Diagnosis and treatment in physical therapy: an investigation of their relationship. Phys Ther 1993;73:568–77.
17. Knutsson E, Richards C. Different types of disturbed motor control in gait of hemiparetic patients. Brain 1979;102:405–30.
18. Delitto A, Cibulka MT, Erhard RE, Bowling RW, Tenhula JA. Evidence for use of an extensor mobilization category in acute low back pain syndrome. A prescriptive validation pilot study. Phys Ther 1993;73:216–22.
19. Richards CL, Mouluin F, Duman F, et al. Gait velocity as an outcome measure of locomotor recovery after stroke. In: Craik RL, Oatis CA, editors. Gait analysis: theory and application. St. Louis: Mosby-Year Book, Inc.; 1995. p. 355–64.
20. Sinkjaer T, Arendt-Nielsen L. Knee stability and muscle coordination in patients with anterior cruciate ligament

injuries: an electromyographic approach. J Electromyograph Kinesiol 1991;1:209–17.

21. Williams KR, Cavanagh PR. Relationship between distance running mechanisms, running economy and performance. J Appl Physiol 1987;63:1236–45.
22. Sutherland DH, Cooper L, Daniel D. The role of the ankle plantar flexors in normal walking. J Bone Joint Surg Am 1980;62A:354–63.
23. Berchuck M, Andriacchi TP, Bach BR, Reider B. Gait adaptations by patients who have a deficient anterior cruciate ligament. J Bone Joint Surg Am 1990;72A:871–7.
24. Kadaba MP, Ramakrishnan HK, Gainey JC, et al. Gait adaptation patterns in patients with ACL deficiency. Trans Orthop Res Soc 1993;18(2):361–77.
25. Patla A. A framework for understanding mobility problems of the elderly. In: Craik RL, Oatis CA, editors. Gait analysis: theory and application. St. Louis: Mosby-Year Book, Inc.; 1995. p. 436–49.
26. Sutherland DH, Kaufman KR. Human motion analysis and pediatric orthopedics. In: Harris GF, Smith PA, editors. Human motion analysis, current applications and future directions. Piscataway NJ: IEEE Press; 1996. p. 219–54.
27. Albany K. Physical and occupational therapy considerations in adult patients receiving Botulinum toxin injections for spasticity. Muscle Nerve 1997;Supp 6:S221–31.
28. Wills CA, Hoffer MM, Perry J. A comparison of foot-switch and EMG analysis of varus deformities of the feet of children with cerebral palsy. Dev Med Child Neurol 1988;30:227–31.
29. Halstead LS, Rossi CD. Post-polio syndrome: clinical experience with 132 consecutive outpatients. In: Halstead LS, Wiechers DO, editors. Research and clinical aspects of the late effects of poliomyelitis. White Plains, NY: March of Dimes Birth Defects Foundation; 1987. p. 13–26.
30. Boda WL, Natelson BH, Sisto SA, Tapp WN. Gait abnormalities in chronic fatigue syndrome. J Neurol Sci 1995;131:156–61.
31. Burgess EM, Hittenberger DA, Forsgren SM, Lindh DV. The Seattle prosthetic foot—a design for active sports: preliminary studies. Orthot Prosthet 1993;37(1):25–31.
32. Gitter A, Czerniecki JM, DeGroot DM. Biomechanical analysis of the influence of prosthetic feet on below-knee amputee walking. Am J Phys Med Rehabil 1991;70:142–8.
33. Torburn L, Perry J, Ayyappa E, Shanfield SL. Below-knee amputee gait with dynamic elastic response prosthetic feet: a pilot study. J Rehabil Res Dev 1990;27(4):369–84.

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SECTION FOUR

Future Directions in Gait Analysis

by Kenton R. Kaufman, PhD, PE

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INTRODUCTION

Gait analysis has advanced considerably over the past century. Since the pioneering work of Braune and Fisher (1), much effort has been put into developing the needed technology for human movement analysis. Automated movement tracking systems have replaced hand digitization. Advances in the aerospace industry have been utilized for the development of force plates for kinetic analysis. Computerized electromyography (EMG) systems have replaced hand palpitation. Currently, the technology and knowledge for gait analysis have advanced to a level that permits rapid analysis.

During the past decade, health care delivery systems have evolved at a pace that few expected. The most visible change is the development of managed care delivery systems. An increasing emphasis is being placed on determining the outcome of various clinical procedures. A number of approaches and methods are applied by doctors, nurses, therapists, and other specialists to prevent a particular condition, ameliorate its effects, or change a given state. A scientific basis for clinical practice is being requested. An increasing emphasis is being placed on obtaining accurate measures to determine the outcome of various clinical procedures.

Gait laboratories can play a key role in these managed care scenarios. The objective measurements provided by gait analysis techniques are central to measurement of the patient's progress. The future of gait analysis will depend upon advances made in experimental, analytical, and interpretation techniques for gait studies.

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HEALTH CARE REFORM

Health care costs have risen dramatically over the last 3 decades (**Figure 1**). In 1960, the national health expenditure for health care was 5.3 percent of the gross national product (2); by 1994, health care expenditure had risen to 13.7 percent. The United States currently spends almost three times more on health care than on education or defense, as can be seen in **Figure 2** (3). The U.S. health care system is the most expensive in the world (**Figure 3**). Health care expenditure in the United States was approximately $950 billion in 1994 (2); this is one-third more than any other industrialized nation. In addition to being the most expensive system in the world, U.S. health care costs are growing more rapidly than those in any other industrialized nation. Personal expenditures from medical care have increased over three-fold in the past six decades (2). Initially, most of the payment for personal health care was from out-of-pocket payments (**Figure 4**), but this percentage has decreased and the contribution from health insurance has continued to increase throughout the decades.

Given these economic realities, much emphasis is placed on systems for delivery of health care. These delivery systems were set up to offer the potential for controlling cost by providing coherent networks to obtain discount pricing by integrating the financing and delivery of medical care. Managed care, and everything that it represents (cost containment, competition among providers, constraints on health services,

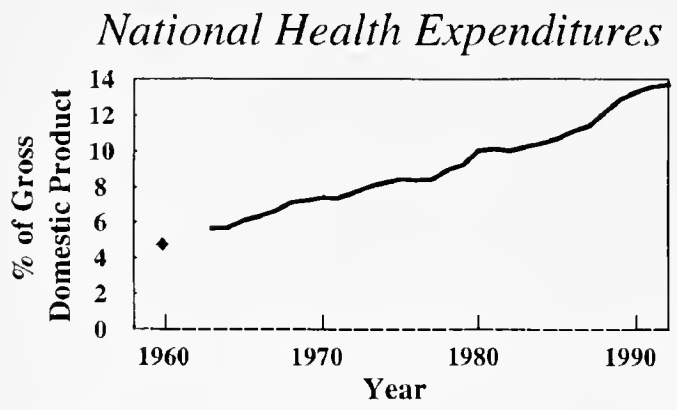

Figure 1.
National expenditures for health care over the last 3 decades, expressed as % of Gross Domestic Product (2).

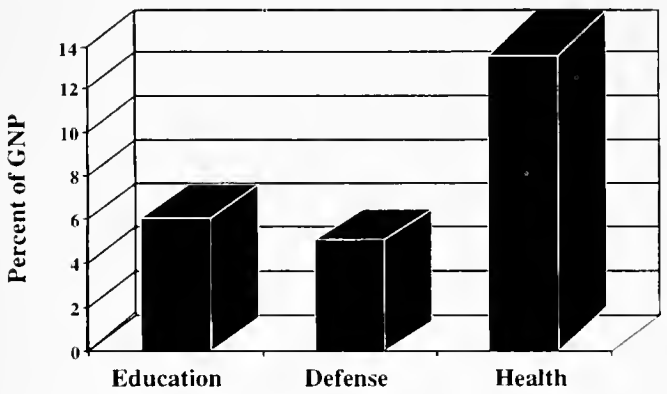

Figure 2.
National expenditures for education, defense, and health in 1994 (3).

reimbursement decreases, and utilization review) has forever changed the traditional fee-for-service model (4). Enrollment in managed health care plans has increased dramatically in the last decade (5). In 1984, 89 percent of patients were covered by an indemnity (fee-for-service) program while less than 10 percent were covered by a managed care program. In sharp contrast, in 1997, 10 percent were covered by an indemnity program while 80 percent were covered in a managed care program. Thus, it is clear that a paradigm shift in health care delivery is occurring.

In the U.S., the evolution of the health care environment varies by location, with the evolution of the market occurring most rapidly in the western states (6). Cost and quality vary widely. Proponents of managed care have shown that this new model for health care delivery has lowered costs, hospital-stays, and mortality rates. Markets with heavily managed care penetration demonstrated that the average hospital costs were reduced by 11.5 percent compared with the national average (5). This was a combined result of providers utilizing fewer resources due to financial incentives and HMOs driving payment rates down. In these highly competitive markets, the average length of stay was also 16.9 percent below the national expected level (5). In addition, hospitals in high managed care markets experienced actual death rates (adjusted for the clinical condition of the patient) that were 8 percent lower than expected on a national basis (5). Nevertheless, one cannot overlook the fact that rationing of "non-medically necessary care" is also being done for financial gain. The proliferation of "for-profit" HMOs has changed the face of health care. Instead of accumulating savings in order to provide better care, these savings go into the pockets of investors in for-profit HMOs. With each year, HMOs have continued to increase their profitability. In 1994, the annual pretaxed earnings of the HMO industry were $4.13 billion (7). Thus, while it is stated that the most significant change in health care has been the shift of risk from payers of health care to the providers, it is also important to note that this shifting of risk can also be related to the shifting of profitability of health care.

It is important that individuals involved in health care policy development, organization, and delivery understand that gait analysis can be used to eliminate unnecessary surgery. DeLuca and colleagues (8) have shown that frequently the number of surgical procedures are reduced after a three-dimensional (3-D) gait analysis, when compared to a clinical examination and videotaping alone. Further, gait analysis will maximize the return when surgery is indicated by providing recommendations for multilevel surgery (9). The use of an appropriately timed gait study should make it possible to develop a treatment plan for a person that can be completed in one operative setting. Objective gait analysis data can be used to quantify the person's functional status; depending on his or her functional status, bilateral multilevel surgery can be performed. When appropriately planned, no further surgery will be needed. This reduction in the number of surgeries will lower the overall long-term cost of treatment. One of the important features of the new health care system will be a greater emphasis on the prevention of disease and measurement of clinical outcome. Patient functional status before and after treatment will need to be studied.

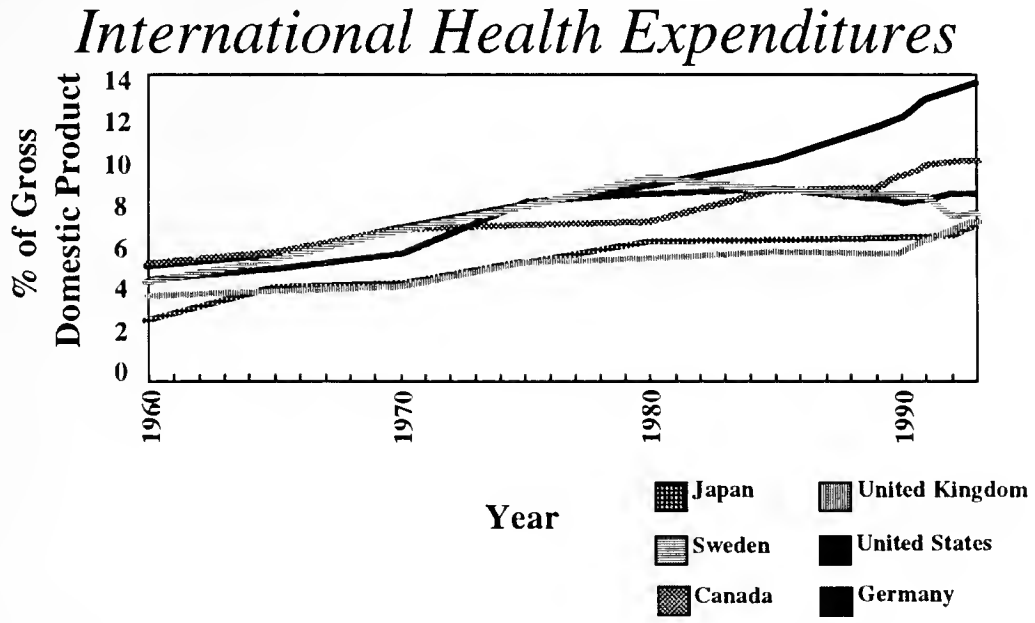

Figure 3.
International expenditures for health care over the last 3 decades, expressed as % of Gross Domestic Product (2).

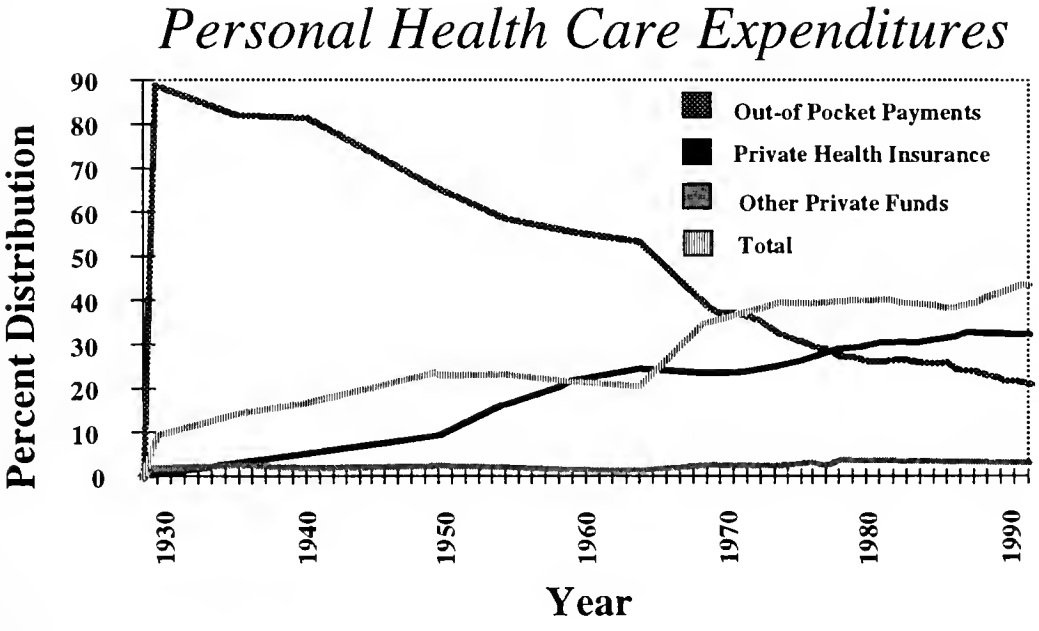

Figure 4.
Personal expenditures for health care over the past 6 decades. The distribution is subdivided into out-of-pocket payments, government payments, private health insurance, and other private funds (2).

Payors are turning their attention from short-term savings to long-term predictable improvement in both cost and quality.

Challenges exist to further evolve the science of clinical gait analysis to make it effective as an assessment tool. Currently, a clinical gait analysis study consists of five broad areas (**Figure 5**). Initially, a history and physical examination is performed on the patient. If it is determined that he or she could benefit from a gait analysis study, an evaluation is requested. The gait study consists of data collection, data reduction, analysis, and interpretation of the results of the study. This information is compiled in a clinical report and recommendation is given for treatment. The future of gait analysis will depend upon advances made in experimental, analytical, and interpretation techniques. Opportunities for future enhancements of gait studies are outlined in the following sections.

EXPERIMENTAL TECHNIQUES

Advances in experimental techniques have been made possible because of the advances in computer technology, which need to be applied toward enhancements of data collection, data presentation, and quantification of muscle function.

Advances in Computer Power

Decreasing costs and simultaneous increases in the computational capacity of computers have facilitated many technological advances in scientific fields. In 1937, Howard H. Aiken of Harvard University conceived the first large-scale automatic digital computer. In the late 1960s, computers operated at an internal speed about 20 to 100 times faster than their counterparts of 10 years earlier. By the 1980s, speeds were 1,000 times faster than in the 1960s. Over the same period, storage capacities and computer memory increased by comparable factors (10). Thus, since 1945 the speed of computers has approximately doubled every 2 years. The exponential increases in the computational power of computers makes development and visualization of biomechanical models of the musculoskeletal system possible. The evolution of performance of microcomputers has surpassed the evolution of conventional supercomputers (**Figure 6**)*. The supercomputer curve shows a steady gradual increase in performance over the last 15 years. In contrast, dramatic improvements in integrated circuit technology are allowing microprocessors to close the performance gap with conventional supercomputers.

Beyond the purely technological improvements in memory and speed, user interface improvements have had an equally large effect on increases in productivity. The interface between the human and the computer has become easier to use and much more efficient. Computer programs of today feature pull-down menus, mouse-driven applications, and graphical input and display capabilities. These changes have resulted in user friendly systems that aid in the visualization and understanding of complicated biomechanical models.

Data Acquisition Systems for Movement Analysis

The techniques for motion analysis have progressed from motion photography, (11,12) and electrogoniometry (13,14) to automated stereometric systems (15–18). Motion data provides the information necessary for calculation of the time/distance parameters of walking (velocity, cadence, stance and swing times, etc.) and the angular position of the person's joints (hips, knees, and ankles) during the different phases of gait. The derived measurements indicate the degree of normalcy (or abnormality) and the presence of compensatory patterns. Techniques that quantify deviation from normal walking during the gait cycle are of greatest clinical interest and treatment potential for people with movement disabilities.

A number of technologies are in use today for the capture of human motion. Each of the existing approaches has undesirable constraints that limit its applicability for real-time modeling of full body motion with the prerequisite for accuracy, scan rate, number of sensors, and range, and without impingement of the limited function of disabled individuals. The five existing technology categories include the

- electromechanical linkage method
- stereometric method
- roentgenographic method
- accelerometric method
- magnetic coupling method.

Each technique has disadvantages.

An exoskeleton apparatus is employed with the electromechanical linkage method to measure joint motion. The primary disadvantage for this technique is the cumbersome nature of the instrument and, to a lesser

*Personal Communication with Lawrence Livermore National Laboratory, Livermore, CA, 1995.

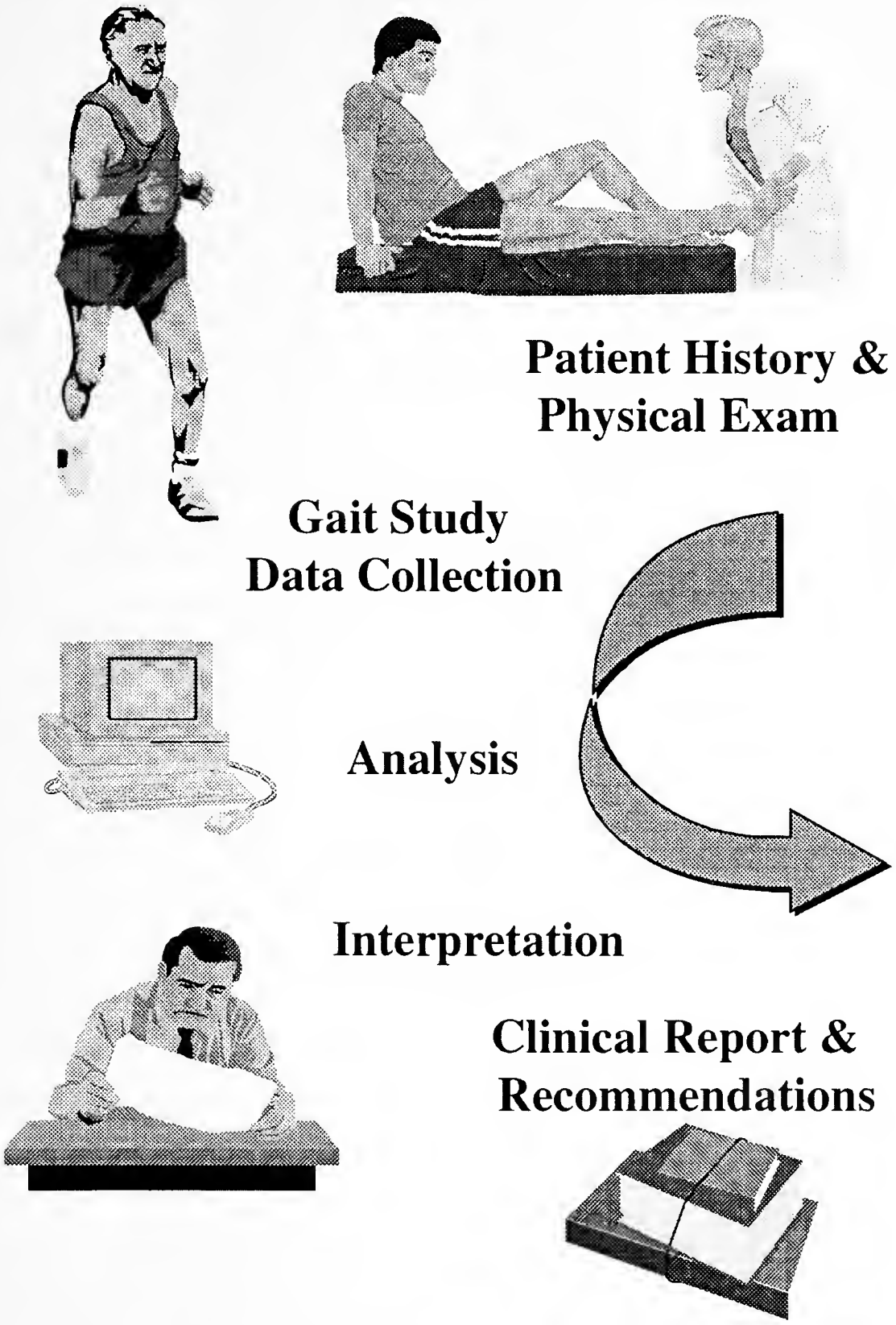

Figure 5.
Sequence of events for a clinical gait analysis study.

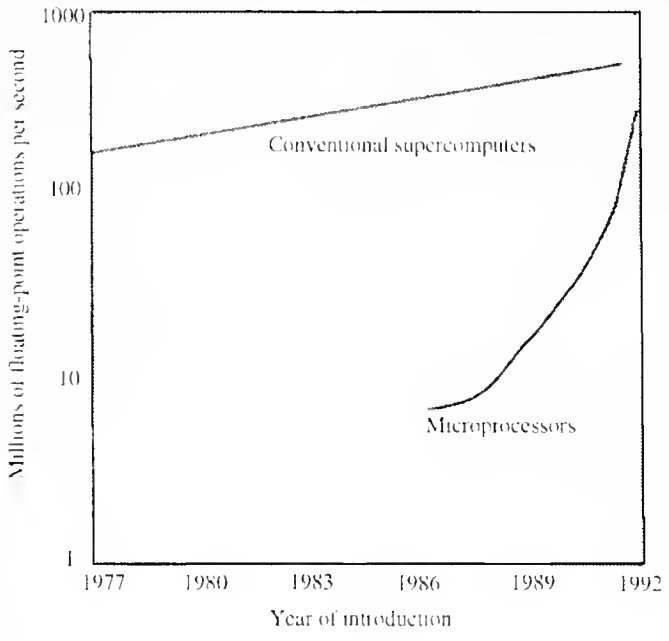

Figure 6.
Evolution of performance of supercomputers and microcomputers (11).

extent, cross coupling of the sensor inputs and joint motion. The requirement for the exoskeleton instrument affects the motion of young subjects making it unusable for clinical measurement.

The stereometric method is the most popular one currently used for clinical gait analysis. It employs visible markers attached to the skin on rigid segments of the body structure and tracks their motion using imaging equipment. This technique is implemented using charge coupled device (CCD) cameras and frame-grabber electronics to allow digital images to be captured as the subject moves within the field of view. Digital image analysis allows the physical location of each marker to be computed, using triangulation of the views from an array of camera systems. This technique has minimal impact on the natural motion of the subject and allows data capture without the need to tether the subject to the data acquisition hardware. However, a disadvantage of this approach is the increased image analysis complexity resulting from tracking the apparent position of the markers in a two-dimensional (2-D) image on a per camera frame-to-frame basis and correlating the position of each marker for the multiple camera positions. Occlusion of markers from the camera field of view and false readings caused by reflection phantoms pose non-trivial, unresolved complications in data capture. In addition, passive markers provide unlabeled trajectory segments that must be manually identified and resolved. This image analysis task requires a significant amount of time for the data gathering process. A second major disadvantage is the reduction in resolution as the camera system is altered to allow a larger field of view. The camera imaging sensors have a fixed number of pixel elements and a compromise must be reached between optical field of view and pixel element resolution size, limiting the clinical measurement volume to approximately a single stride. It is not feasible to measure gait patterns or variability with only one traversal of the instrument walkway. Thus, multiple walking trials need to be collected, which may fatigue the subject.

The biplanar roentgenographic method employs metal markers and x-ray films for the measurement of static positions of a body joint. This approach is not appropriate for the study of dynamic joint motion. Due to the use of ionizing radiation, it also represents a potential health hazard to the subject.

The accelerometric approach employs sensors attached to the rigid areas of the human subject that measure accelerations in three dimensions. Joint motion is then derived through integration of the accelerometer waveforms given appropriate initial conditions. Integration of the waveforms produces velocities for each of the sensor locations. A second integration step provides the displacement as a function of time. This technique can provide the kinematic motion measurement desired but has been implemented with a tether to the subject for the data acquisition; however, the tether affects the motion of the subject and represents an undesirable feature. In addition, this approach requires an accurate estimate of initial conditions, which is difficult to provide.

The magnetic coupling method employs a reference magnetic field source that surrounds the subject with an array of magnetic field sensing elements attached to the rigid segments of the subject. The position of each sensor is estimated through analysis of the magnetic field components passing through the sensor. This technique has the potential for providing complete six degree-of-freedom motion information, but has been implemented by using a tether to each of the sensors and the data collection system. The response of the system is limited to 30 Hz, which is below the required data acquisition rate for high fidelity measurements. Further, the sensing elements are sensitive to

nearby ferromagnetic materials that may distort the field.

While the foregoing techniques have provided a means for the acquisition of joint motion, there are associated deficiencies for each. A system needs to be developed for real-time acquisition of human motion. The goal of this effort should be the development of a technique for precise measurement of human body motion. Suggested guidelines for the performance of the proposed system are given in **Table 1.** The unique characteristics specified request that the real-time system be able to function over a larger measurement area using high-scan rates and a large number of fiducial points. These system requirements will lead to greater accuracy than that which exists in current systems. These enhanced capabilities will eliminate clustering limitations and data reduction, reduce the cost of gait analysis as a clinical treatment planning service, and improve the turnaround and availability of information for clinical decision making. The advantages to be realized include a real-time motion acquisition and display, higher data sample rates, substantially increased work volumes, full body motion acquisition, reduced data loss from occlusion, and a significant time savings for data analysis. This development will open new windows of opportunity for the application of motion analysis to sports injuries and other domains requiring higher scan rates and larger measurement volumes. Successful development and commercialization of a real-time data acquisition system represents a new paradigm for human movement analysis.

Table 1.
Specifications for a real-time motion system.

Characteristic	Specification
Motion	Reported with respect to an absolute reference frame in real time
Sampling Rate	60 scans/s min, 200 scans/s desired
Resolution	1 mm
Latency	500 ms max
Limb Obstruction	None
Marker Numbers	30 min, 300 max
Marker Separation	1 cm max
Capture Range	0.5 m min, 50 m max
Subject Movement	Unaffected by instrumentation

min = minimum; max = maximum

Visualization of Human Motion

Gait analysis typically includes measurements of motion, force, and muscle activation patterns (electromyography). In recent years, dynamic measurements of foot pressure have been added to the armamentarium of diagnostic tools. The clinical interpretation of pathological gait requires holding in human memory a large number of graphs, numbers, and clinical tests from data presented on hard copy charts, x-rays, video, and computer-generated 3-D graphics from multiple trials of a subject (**Figure 7**). Further, comparisons must be made to data from a normal population in order to identify the potential movement problems for a given individual. The referring physician, who is not an expert in gait analysis, is overwhelmed by the magnitude of the number of measurements included in a typical clinical report. This information must be integrated into a cohesive plan for clinical intervention, which often includes multiple surgeries. While data collection techniques for gait analysis have continually evolved over the past 40 years, the method of data presentation has not changed over this same time. The data is still reported in 2-D charts with the abscissa usually defined as the percentage of the gait cycle and the ordinate displaying the gait parameter.

Recent developments in computer animation make it possible to apply advanced methods to visualize human movements. A highly dimensional space is needed to fully describe the complexities of human movement. The large volume of variables currently found in a typical clinical report should be replaced with a printout of a few graphic images that succinctly provide the needed information. It is difficult to fully appreciate and understand relationships between motion dynamics and physiologic or biomechanical variables without scientific graphic visualization.

Due to the complexity of gait-derived data, powerful visualization tools are needed. The ability to incorporate scientific visualization will provide unprecedented power to support the clinician's recommendations in a manner that the referring physician can intuitively understand and visualize. The popular scientific visualization techniques are 1) one-dimensional (1-D) plotting, 2) 2-D plotting, 3) 3-D volume visualization, 4) imaging processing, and 5) animation (19). Separate software packages are available to perform each of these techniques. However, as the need to solve

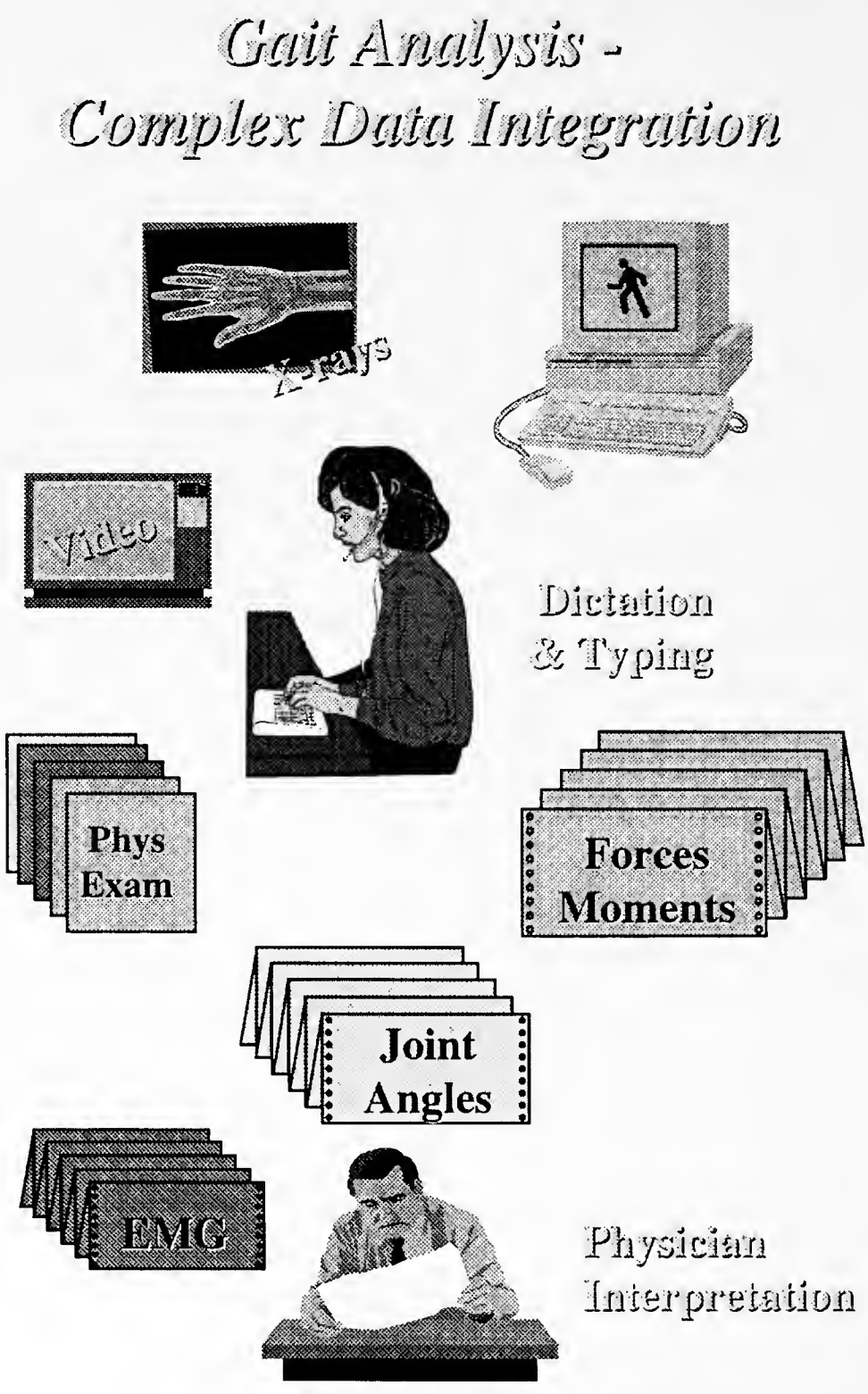

Figure 7.
Current situation for clinical interpretation of gait analysis reports. The large number of graphs, numbers, and clinical tests must be held in human memory in order for the appropriate comparisons to be made.

complex problems becomes more acute, one package is needed that provides all of these capabilities to enhance productivity.

A scientific computing environment is needed that will allow the rapid transmission, archival, retrieval, and manipulation of images within a system equipped with analytical tools useful for clinical and research purposes. Tools are needed for data collection, analysis, and visualization (**Figure 8**). A suitable database of normal gait patterns is needed for comparison. The ultimate goal of this scientific visualization workstation is to provide a user-friendly, menu-driven environment that will facilitate the reporting of biomechanical data and integrate real-time animation of fully 3-D realistic graphical depictions of articulated body segments. This system should provide clinicians with the ability to visualize the correlation between collected biomechanical data and the actual human motion. Furthermore, this system should provide the ability to simulate gait and compare the computer-generated simulation with experimentally collected data. The operator should be able to examine the data from any viewing angle, to zoom in or out, change the viewing perspective, or stop the motion. This system should have the ability to superimpose normal gait on a subject's gait in order to visualize differences. The system should also be able to align the bodies displayed to a common center of gravity or to a common point in the gait cycle (20). It should be capable of "removing" extremities in order to improve visualization of other body segments. The two most important goals are the realistic appearance of the human figure and the convenient specification of the biomechanical data. This application should be user-friendly so that it can be used by colleagues who are not necessarily programmers but have the expertise in their respective fields (e.g., medical doctors). The software environment should be capable of quick and easy customization to serve very specific needs.

Another key issue is the communication of the results. The clinician must be able to select only the most essential results for the communication to the referring physician and the patient. Otherwise, the individuals will be overwhelmed by the plethora of numbers while comprehending little. Practical display of data will provide an economical and efficient method of communicating information (21). When graphical portrayal of data is limited by dimensionality (i.e., three dimensions), other variations in the output such as color, sound, and shape can be used to help overcome this limitation. This application of technology should provide a mechanism to integrate all aspects of gait measurements and observations into a single tool for physician interpretation, diagnosis, and treatment recommendations (**Figure 9**).

Muscle Force Measurement

Muscle forces reflect the underlying neurocontrol processes responsible for observed movement patterns. In addition, muscle forces play a major role in determining stresses in bones and joints. Thus, a knowledge of muscle forces is fundamental for improving the diagnosis and treatment of individuals with movement disorders. Interpretation of muscle function has routinely been based on analyses of electromyographic data obtained during gait studies (22–24). More specific detailed knowledge of the muscle forces acting on the body will allow us to improve our ability to diagnose and treat persons with movement disabilities. It will also increase our understanding of muscle function during gait. Unfortunately, invasive techniques for measuring muscle forces are highly objectionable.

Techniques such as electromyography (EMG) do not provide the quantitative accuracy needed. A fundamental relationship exists between the tension that a muscle is capable of developing and the length of the muscle. The total muscle tension is composed of both active and passive components. This well-known phenomenon is described by Blix's curve, which demonstrates the relationship of total muscle force, passive stretch force, and muscle contractile force to the length of the muscle (25). Yet, the integrated electromyogram can only be proportional to the active component and will not account for the passive stretch of the muscle. Use of the integrated electromyogram as an indicator of the quantity of muscle contraction has another drawback. There is a significant delay between maximal electric activity in the muscle and maximal tension. The electromechanical delay has been estimated to be 30 to 90 ms (26–30), which would be approximately 3 to 9 percent of the gait cycle.

Measurement of intramuscular pressure is a conceivable solution. Intramuscular pressure is a mechanical variable that is proportional to muscle tension. Investigators have shown in studies on animals (31–33) and in humans (34–36) an approximately linear relationship between intramuscular pressure and muscle force during isometric muscle contraction. Further, estimation of muscle force from intramuscular pressure is not affected by changes in signal due to muscle fatigue

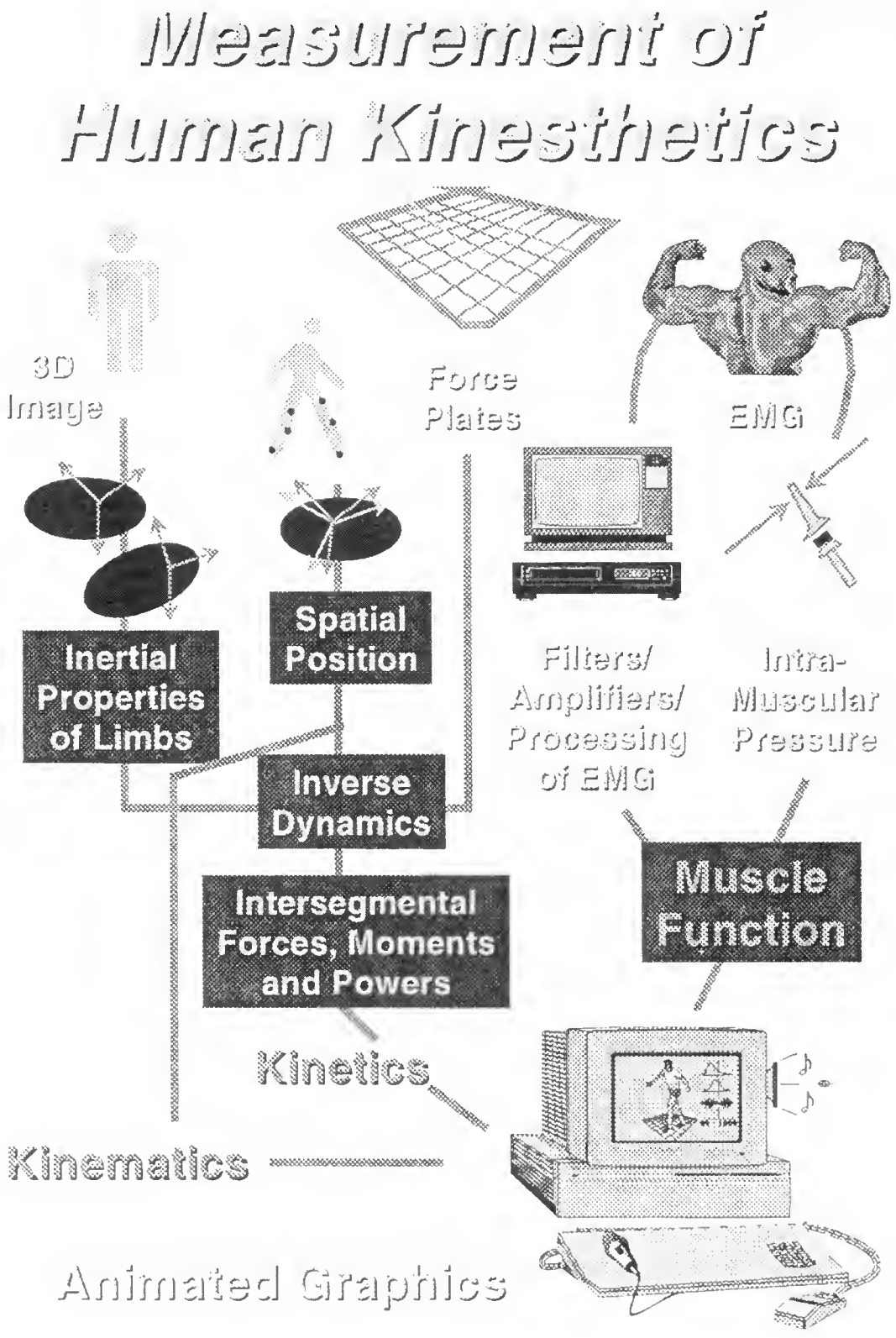

Figure 8.
Scientific computing environment needed for collection and visualization of human kinesthetics.

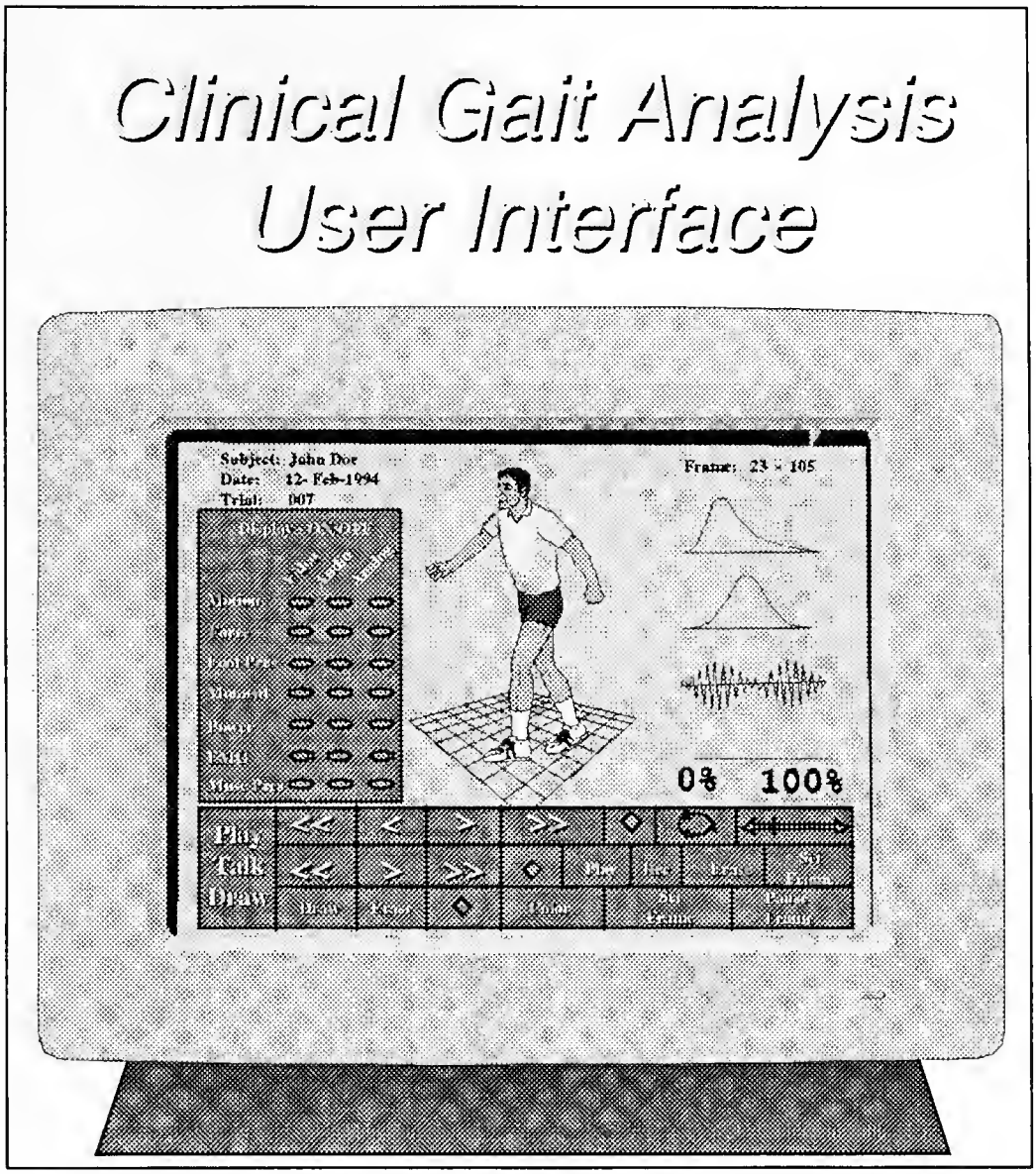

Figure 9.
User interface for graphical portrayal of gait analysis data.

(37,38). Nevertheless, the absolute intramuscular pressure depends on the depth of the recording catheter within the muscle (32,38), the shape of the muscle (39), and the compliance of the surrounding tissue (40). Baumann et al. (41) reported that intramuscular pressure is related to the active and passive components of muscle tension during gait. Kaufman and Sutherland (42) have also reported that the intramuscular pressure during walking parallels the electromyographic activity, but also accounts for passive stretch of the muscle (**Figure 10**). In the future, more work is needed in the use of intramuscular pressure to quantify muscle force. Improvements in microsensor technology can be used to facilitate these measurements.

ANALYTICAL TECHNIQUES

A fundamental concern in the study of human locomotion is a description of the kinematics and kinetics involved. During the study of gait, a large number of measurements are taken. The experimental data are entered into an analytical model to obtain the values of variables not directly measurable. The analytical model is a link segment model. The human body is modeled as a system of articulated, rigid links, which represent the lower limb segments and the upper body. By modeling the body as an ensemble of rigid-body segments, it is possible to calculate the movement and loads at any articulation.

Kinematics

In order to establish a mathematically workable model, Cartesian coordinate systems are established on each body segment (43–47). These anatomically based axis systems are fixed in each body segment and move with it. The coordinates of bony landmarks are used to build a right-handed orthogonal coordinate system. The unique specification of anatomical coordinate systems requires a minimum of three noncolinear points that are defined with respect to surface landmarks associated with each segment. In order to obtain the joint movement, expressions have to be obtained relating the position of each segment in the model with respect to adjacent segments. Joint motions are usually 3D. The anatomical description of the relative orientation of the two limb segments can be conveniently obtained by relating the two coordinate systems embedded in the proximal and distal body segments. The ability to describe joint orientation in 3-D space following traditional rigid-body motion theory is essential. For finite spatial rotation, the sequence of rotation is extremely important and must be specified for a unique description of joint motion. For the same amount of rotation, different final orientations will result from different sequences of rotation. However, with proper selection and definition of the axes of rotation between two bony segments, it is possible to make the finite rotation sequence independent or commutative. In the past 15 years, the concept of Eulerian angles has been adopted in the field of biomechanics to unify the definition of finite spatial rotation (48,49). If a unit vector triad (**I,J,K**) is attached to a fixed segment along the XYZ axes and another triad (**i,j,k**) is fixed to the moving segment along the xyz axes (**Figure 11**), the relationship between them after any arbitrary finite

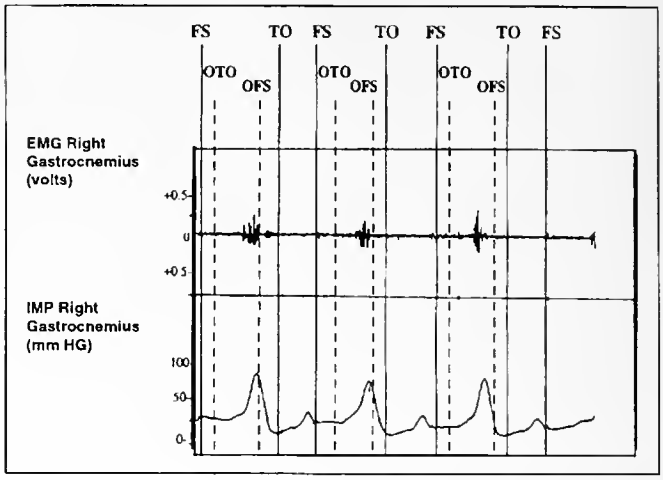

Figure 10.
Raw data for single subject during gait: EMG and intramuscular pressure are being recorded from the gastrocnemius muscle. Stance phase of gait occurs from FS to TO. Swing phase of gait occurs from TO to FS and single-limb stance from OTO to OFS. Peaks in intramuscular pressure during gait can be correlated with peaks of active contraction and passive stretch of the gastrocnemius (42).

rotation can be expressed by a rotational matrix in terms of three Eulerian angles, ϕ, θ, ψ as follows:

$$\begin{bmatrix} \mathbf{i} \\ \mathbf{j} \\ \mathbf{k} \end{bmatrix} = \begin{bmatrix} c\theta c\phi & c\theta c\phi & -s\theta \\ -c\psi s\phi + s\psi s\theta c\phi & c\psi c\phi + s\psi s\theta s\phi & s\psi c\theta \\ s\phi s\psi + c\psi s\theta c\phi & -s\psi c\phi + c\psi s\theta s\phi & c\psi c\theta \end{bmatrix} \begin{bmatrix} \mathbf{I} \\ \mathbf{J} \\ \mathbf{K} \end{bmatrix}$$

where s=sine and c=cosine. The Eulerian angles can be calculated based on the known orientation of these unit vector triads attached to the proximal and distal body segments.

For a more general unconstrained movement in space, three translations and three rotations are required to describe the joint motion. The displacement of a rigid body can take place along any one of an infinite number of paths. It is convenient to describe the displacement in terms of the simplest motion that can produce it. The most commonly used analytic method for the description of six-degree-of-freedom rigid-body displacement is the screw displacement axis (50–52). The motion of the moving segment from one position to another can be defined in terms of a simultaneous rotation, Φ, around and a translation, τ, along a unique axis, called a screw displacement axis, which is fixed in the fixed segment (**Figure 12**). The screw displacement axis is a true vector quantity. However, the amount of the finite screw rotation is not a vector quantity, and the

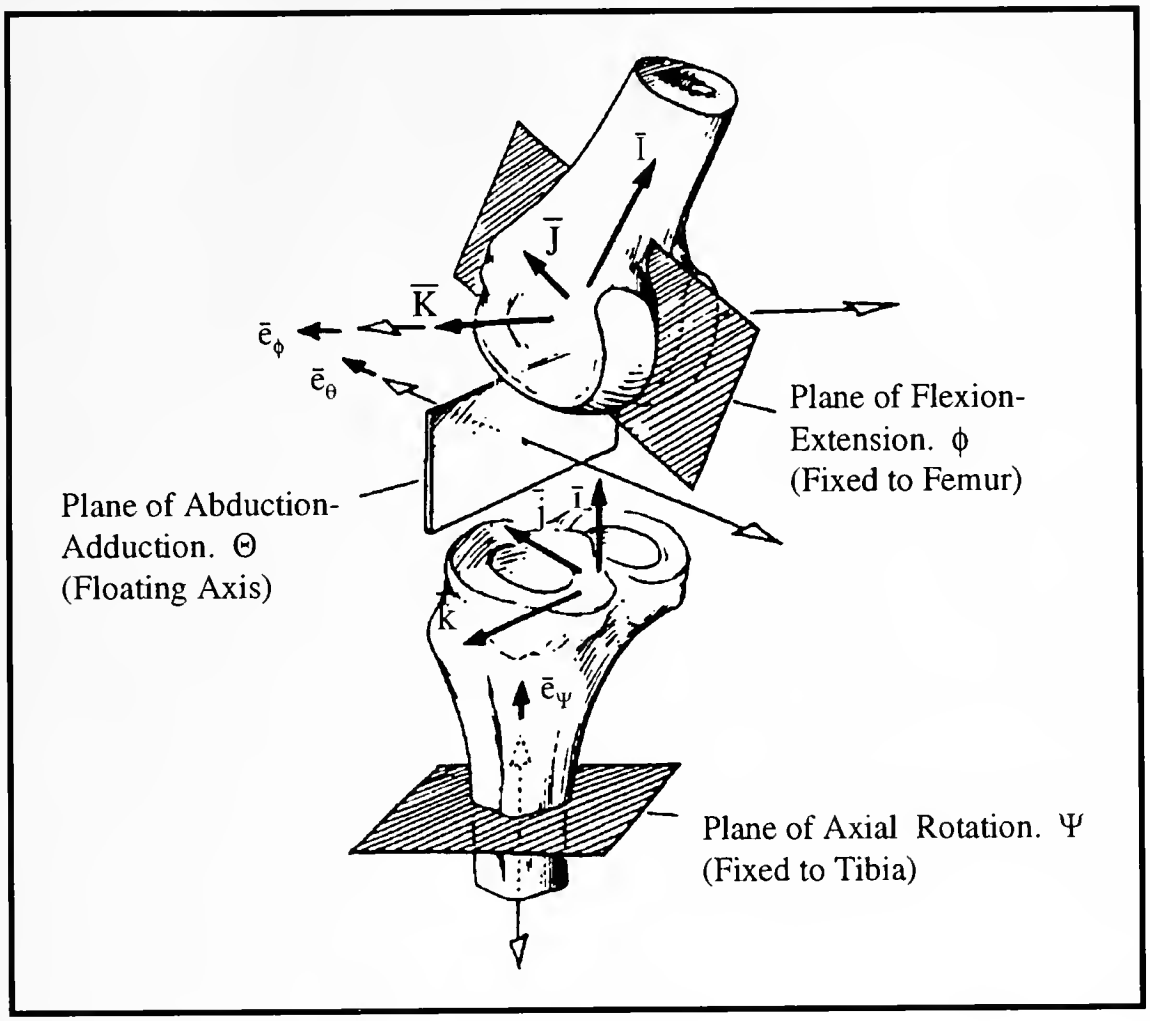

Figure 11.
Description of knee joint motion using Eulerian angle system. Axis fixed to distal femur defines flexion/extension motion, Φ. Axis fixed to proximal tibia along its anatomical axis defines internal-external rotation, ψ. Floating axis is orthogonal to other two axes and used to measure abduction-adduction, θ. (Reproduced with permission of Mayo Foundation.)

decomposition of it must be carefully interpreted because of the noncommutative nature of finite rotation. Woltring (53) recommended that the component rotations (flexion-extension, abduction-adduction, endo-exo-rotation) be defined as a component of the product $\mathbf{\Phi} = \mathbf{\Phi n}$ where **n** is the unit direction vector of the screw axis.

Inverse Dynamics

Once the transformation matrices have been obtained, we can proceed to solve for the joint moments given the joint positions, velocities, and accelerations, and the ground reaction forces. Typically, these formulations are based on the inverse dynamics approach (54), proceeding from known kinematic data and external forces and moments to arrive at expressions of the resultant intersegmental forces and moments. If the exact motion history of the system, especially accelerations, is available, then this type of problem presents little mathematical challenge and can be solved by applying the equations of motion derived for the system (**Figure 13**). An unconstrained rigid body has six

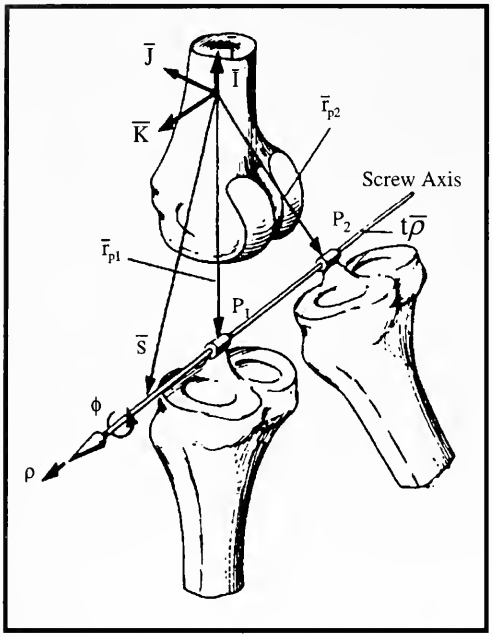

Figure 12.
Screw displacement axis can be used to describe general spatial motion. Tibia moves from position 1 to position 2 by rotation about screw axis by an amount Φ and by translating along the screw axis by an amount τ. (Reproduced with permission of Mayo Foundation.)

degrees of freedom. Hence, six equations of motion are needed to specify its configuration. Three equations can be chosen to represent the translation of the rigid-body center of mass and three equations to represent the rotation about any point, A. In the case of the motion of a rigid body in three dimensions, the fundamental equations are:

$$\Sigma \mathbf{F}_A = m\mathbf{d}(\dot{\mathbf{r}}_c)/dt$$

$$\Sigma \mathbf{M}_A = \dot{\mathbf{H}}_A + m(\dot{\mathbf{r}}_A \times \dot{\mathbf{r}}_c)$$

These fundamental equations express that the system of external forces, $\Sigma \mathbf{F}_A$, and moments, $\Sigma \mathbf{M}_A$, acting at the limb segment are equipollent to the system consisting of the linear momentum vector, m $d(\dot{r}_c)/dt$, and the moment of momentum vector, $\dot{\mathbf{H}}_A + m(\dot{\mathbf{r}}_A \times \dot{\mathbf{r}}_c)$. Using measurements of the intersegmental load actions and the relevant kinematics, it is possible to compute the energy and power transmitted from one body segment to another. The joint powers are obtained from the scalar (dot) product of the intersegmental joint moment and the joint angular velocity as well as the intersegmental joint force and translational velocity. The rate of work done (power) can be calculated from:

$$\dot{W} = \mathbf{M} \cdot \boldsymbol{\omega} + \mathbf{F} \cdot \mathbf{v}$$

where $\dot{W}$ = mechanical power
$\mathbf{M}$ = intersegmental joint moment
$\mathbf{F}$ = intersegmental joint force
$\boldsymbol{\omega}$ = angular velocity and
v = translational velocity.

Frequently, the component due to translation is assumed to be small and the second term (F· v) is rarely included in joint power estimates for gait. This technique can be used to predict the transfer of energy from body segment to body segment through the muscles (55). The muscles can either generate or absorb mechanical energy by contracting concentrically or eccentrically, respectively.

Body Segment Mass Inertial Estimates

Estimates of body segment mass, center of mass, and moments of inertia are needed for these biomechanical models. These body segment parameters are used along with the segmental kinematics to compute the linear and angular momentum of the body segments. Estimates of these values are substituted into the Newton-Euler equations of motion to obtain an estimate of joint loads during physical activity. These body segment estimates are a big source of error in biomechanical models (56). Methods of obtaining inertial parameters of body segments can be classified into three groups:

- *regression equations*
- *geometric approximation*
- *direct measurement*

Regression equations have been developed based on cadaver studies (57–59) and living subjects (60–62). The equations have been developed through statistical analysis of the data. The regression equations based on cadaver studies typically lead to errors arising from differences in tissue composition and morphology between the cadaver samples and a given human subject (63). The study by Chandler et al. (59) was the first study to determine the segmental principal axes of

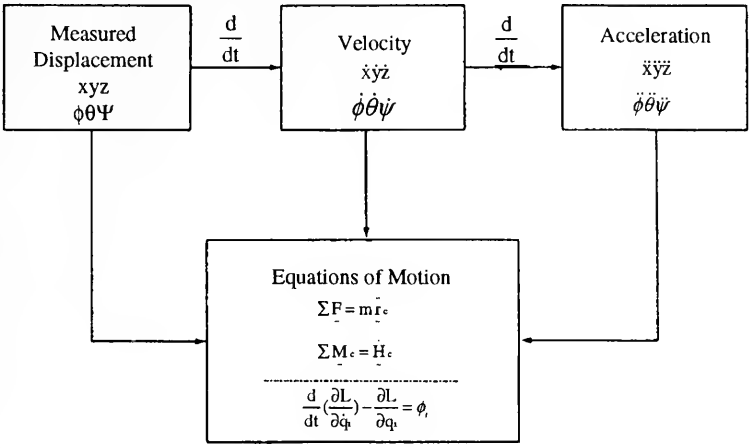

Figure 13.
Solution process for inverse dynamics problem. Displacement information must be differentiated twice to yield acceleration. Either Newtonian or Lagranian formulations can be used to formulate the equations of motion. (Reproduced with permission of Mayo Foundation.)

inertia and provided verifiable comparisons of derived photometric values and directly measured values. On the basis of these comparative relationships, a series of predicted regression equations were developed for adult males (61) and adult females (62). However, the sample sizes of these studies have been relatively small.

The *geometrical approximation* method represents the shape of different body segments with standard geometric forms that are capable of simple mathematical description (64–68) or magnetic resonance imaging (69,70). However, such techniques can involve high radiation levels (computed tomography) and require specialized, expensive instrumentation.

Future work should be aimed at obtaining inexpensive, fast, noninvasive, individualized estimates of the inertial properties of body segments based on *direct measurements.* One possibility is a video-based system (68). Error levels using this technique are on the order of 5 percent. Another possibility is the use of high-speed laser scanning. A 3-D laser scanner can obtain digitized images of a subject's limb in 10 seconds (71,72). Markers placed near anatomic landmarks can be used as reference points. These data can be used to compute subject-specific body segment parameters.

Forward Dynamics

These biomechanical models of the musculoskeletal system have improved our understanding of the complex processes underlying movement. Traditional gait studies have typically been conducted to collect experimental data and analyze movement and forces. In the future, the forward dynamics model can be used more extensively to study how the body actually produces movement. The forward dynamics problem provides the motion of a multibody system over a given time period as a consequence of the applied forces and given initial conditions. Solution of the forward dynamics problem makes it possible to simulate and predict the body segment's motion. The resultant motion is a result of the forces that produce it. Numerical computation of movements produced by applied forces can lead to an improved understanding of the locomotor system.

Using models to synthesize gait can provide insight into the relationship between muscle forces or joint

moments and the body segment motions that result. The equations that govern the motion of the body can be expressed as:

$$[\mathbf{H}(\theta)]\ddot{\theta} = \mathbf{C}(\theta,\dot{\theta}) + \mathbf{G}(\theta) + \mathbf{F}_m(\theta)$$

where [H(θ)] is an n x n inertia matrix for an n degree of freedom model

$\mathbf{C}(\theta,\dot{\theta})$ is an n x 1 vector of coriolis and centrifugal terms

$\mathbf{G}(\theta)$ is an n x 1 vector of gravitational terms

$\mathbf{F}_m$ is an n x 1 vector of applied moments

$\theta,\dot{\theta},\ddot{\theta}$ are all n x 1 vectors of angular displacement, velocity, and acceleration.

Solving directly for the vector of angular acceleration gives:

$$\ddot{\theta}=[\mathbf{H}(\theta)]^{-1}\{\mathbf{C}(\theta,\dot{\theta}) + \mathbf{G}(\theta) + \mathbf{F}_m(\theta)\}$$

Dynamic simulations of movement integrate this equation forward in time to obtain motion trajectories in response to neuromuscular inputs (**Figure 14**). The inputs can be either joint moments or muscle forces that act on the skeletal system to result in joint moments. Experimentally collected kinesiological data (i.e., body segment motion, ground reaction forces, and electromyographic data), can be used to compute the forward dynamics model inputs that give the measured motion trajectories. The simulated gait pattern can be studied to gain insight into the muscle coordination of the task.

Currently available models for simulating human locomotion have tended to be simple (73–83). State-of-the-art mathematical models of the musculoskeletal system need to be developed to predict gait patterns. The forward dynamics problem seeks the solution to a system of nonlinear ordinary differential equations (initial value problem). These differential equations are numerically integrated starting from the initial conditions. An important characteristic of this mathematical problem is that it is computationally intense. Because of this characteristic, it is very important to choose the most efficient method for solving this problem. Mathematical models have not been fully developed for several reasons:

1. The development of a dynamic model of the body that is sufficiently complex to encompass the multijoint, multibody, multimuscle characteristics of the human body requires considerable effort (84). The problem is to develop a model that is phenomenologically correct without being overwhelmingly complex for practical applications.
2. The muscle excitation patterns required as input to such a model are not fully defined (85). An improperly designed neural excitation pattern will simply result in inadequately coordinated body segment displacements.
3. The dynamic optimization algorithms to find iteratively an acceptable muscle excitation pattern are few and lack robustness (86).
4. The computational time required to find an adequate muscle excitation pattern is long (87).
5. The coordination principles provided by the neurological control systems in unimpaired individuals are poorly understood. The additional challenges of understanding pathological neuromuscular control systems have yet to be addressed.

When fully developed, these models need to include representations of the muscle tendon complex (87–91), skeletal geometry (92,93), kinematic models of the anatomic joints (94), and inertial characteristics of the body segments (95). Realistically developed theoretical models of the neuromusculoskeletal system will play a significant role in understanding locomotion.

Computer-based models are needed to study the biomechanical consequences of surgical reconstructions of the lower extremity. Upon review of data from a gait analysis study, surgical reconstruction is frequently

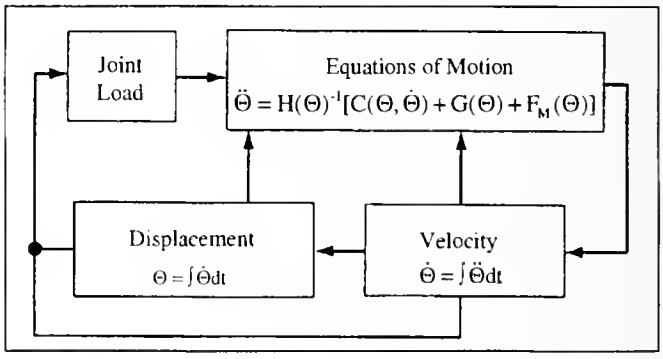

Figure 14.
Solution process for the forward dynamics problem. Joint load can be taken directly from joint moments or can be calculated from a muscle-tendon model and a joint moment arm model that yield joint moments. Joint loads cause angular accelerations, $\ddot{\Theta}$. Equations of motion are integrated to yield joint velocities, $\dot{\Theta}$, and displacements, Θ.

recommended. Sometimes the reconstructive procedure compromises the capacity of muscles to generate forces and moments about the joints. Computer models are needed to predict the anticipated effects that surgical alterations to the musculoskeletal system will have on a person's gait pattern. Relatively few researchers have developed computerized musculoskeletal models to plan orthopedic reconstructive surgeries for correcting pathological gait. Johnson et al. (95) developed a computer model of the hip to evaluate effects of surgical alterations. Dul et al. (96) developed a biomechanical computer model to simulate tendon transfer surgeries to correct equinovarus. Lindgren and Seireg (97) studied the effects of mediolateral deformity, tibial torsion, and different centers of foot support during gait in persons with varus deformity of the knee. Delp et al. (93) developed a graphical model of the lower limb to visualize the musculoskeletal geometry and manipulate model parameters to study the biomechanical consequences of orthopedic surgical procedures. Mann (98) developed a surgical simulation model to determine the effect of skeletal system alterations on subjects' specific gait patterns. Typically these surgical models compute static changes to isometric conditions but do not extend to dynamic movements such as gait. Future work is required to enhance these models.

Future models should include the 3-D characteristics of the musculoskeletal geometry, as well as subjects' specific parameters. The musculo-tendinous aspects of the model need to be scaled to the individual being studied. The numerical ability to predict ambulation following changes to the musculoskeletal system is imminently feasible. However, it has not currently been implemented.

INTERPRETATION OF GAIT STUDIES

Despite the growing availability of technology, gait analysis has not yet become a common tool for the physician. Gait laboratories have been started when individuals and institutions were willing to make the investment in time, effort, and money to assemble and operate gait laboratory systems. Gait laboratories have flourished when a combination of physician input and referral was coupled with day-to-day expertise in the form of physical therapists or other health care specialists and with technical expertise in the form of engineers and other technical staff (99). Several commercial gait systems are on the market. Increasing interest in gait analysis is emerging. Sixty-eight percent of the clinical gait analysis laboratories in the United States have been developed in the last 10 years (100). This trend demonstrates that gait laboratories are becoming recognized as an important clinical tool in the assessment of gait abnormalities.

When new gait laboratories are started, they frequently make a sizable investment in equipment. Nevertheless, instrumentation alone cannot make gait analysis clinically relevant. Clinical gait analysis includes the correlation and interpretation of the data. Taking care of patients in a gait laboratory requires turning data into information. The problem-solving process requires questioning the patient; performing a physical examination; obtaining kinematic, kinetic, and electromyographic data; and linking the symptoms (complaints), signs (physical exam), and test results (gait data) to obtain a treatment plan (**Figure 5**). In this process, it is important to distinguish between functional deficits that contribute to the individual's problem and compensations that the patient adopts in an attempt to walk more normally. The future of gait analysis lies in the ability to process data quickly and identify the patient's functional deficits. Classification methods are needed to characterize a person's gait and direct the clinician reading the gait study to the movement abnormalities. The ability to develop computerized classification techniques will make gait analysis accessible to a wider audience with limited experience. The initial step is to develop standards for collection, reduction, and reporting of clinical gait data.

STANDARDIZATION OF GAIT ANALYSIS TECHNIQUES

Standardization of gait analysis techniques must be established so that data can be shared between laboratories for expert consultation. Several national organizations are undertaking these endeavors. Standardized techniques are being defined for appropriate studies in various clinical settings. Measurement of normal and pathological movement for the purpose of providing recommendations for therapeutic treatment has been successfully achieved by practitioners and laboratories. However, approaches to data collection, reduction, presentation, and interpretation have varied considerably because of differences in equipment, facilities, person-

nel, and philosophy. The result is clusters of methodology without specific guidelines for comparison and communication.

The North American Society for Gait and Clinical Movement Analysis recognizes the need to facilitate communication and encourage the interchange of information among the many professionals who assess the problems of human movement. As a means for reducing confusion, this society has established a Standards Committee to define standards that can be adopted to achieve uniformity in clinical movement analysis. The Standards Committee was formed to achieve standardization of

1. nomenclature use in the collection, reduction, and presentation of data
2. approaches and techniques for data acquisition and reduction
3. quality assurance techniques
4. the form for presentation of results
5. methods for interpretation and reporting clinical findings
6. a format for sharing data between laboratories.

The Standards Committee intends to make contact with and work in concert with any and all parallel bodies that may exist in other specialty societies.

Accreditation is needed to assure quality and achieve continuous improvement of clinical gait and movement analysis. Accreditation attempts to assure that laboratories provide patient care that is effective in contemporary practice. Accreditation will publicly recognize those laboratories demonstrating a higher level of performance, integrity, and quality, which entitles them to complete confidence of the movement analysis profession. Accreditation efforts are occurring at two levels in North America.

The North American Society of Gait and Clinical Movement Analysis has established an Accreditation and Guidelines Committee. The Accreditation and Guidelines Committee will

1. serve as a liaison to nationally recognized accreditation boards pertaining to gait and clinical movement analysis
2. develop and recommend criteria for accreditation that may be used to evaluate the quality of patient care provided by laboratories involved in gait and movement analysis
3. bring together practitioners, evaluators, and administrators in an activity directed toward the continuous development and improvement in the quality of clinical movement analysis throughout North America
4. establish a process for continuous self-study and improvement of movement analysis professionals and laboratories.

The Commission for Motion Laboratory Accreditation has been formed as a non-profit organization. It was developed to enhance the clinical care of persons with disorders of human motion. These goals will be achieved by

1. developing measurement standards to improve the utilization of gait and human motion laboratories for clinical diagnostic purposes
2. evaluating and requiring human motion laboratories to meet a set of standard criteria that will include clinical indication, measurement precision, measurement accuracy, and uniformity of terminology.

The Commission contains representatives from the American Academy of Cerebral Palsy and Developmental Medicine, Pediatric Orthopedic Society of North America, American Society of Biomechanics, American Academy of Orthopedic Surgery, American Academy of Physical Medicine and Rehabilitation, North American Society of Gait and Clinical Movement Analysis, American Orthopedic Foot and Ankle Society, and the American Physical Therapy Association. The Commission will start to accredit laboratories in 1998.

Similar efforts are underway in Europe. The Computer-Aided Movement Analysis in a Rehabilitation Context (CAMARC) project is being undertaken under the Advanced Informatics in Medicine Action of the Commission of the European Communities with academic, industrial, public health, and independent partners from Italy, France, the United Kingdom, and The Netherlands.

The aims of the project are the

1. assessment of existing biomedical knowledge of movement analysis
2. standardization of test protocols
3. assessment and implementation of relevant digital signal processing algorithms
4. analysis of marketing potential of new instrumentation
5. development of design criteria for new devices.

It is the hope of this group to develop standards in the appropriate interface between the instrumentation and a suitable neuromusculoskeletal model. Accommodation of movement data and an appropriate model of human movement are expected to provide meaningful information for assessment of unimpaired and pathological movement for diagnosis, treatment planning, pre- and post-treatment comparison, and long-term follow-up.

CLASSIFICATION TECHNIQUES

One of the main obstacles to automated gait analysis is the difficulty of distinguishing between normal and abnormal movements. A person's gait is classified as abnormal when the person's gait parameters deviate excessively from normal. The clinical application of gait analysis is aimed at identifying these inappropriate deviations. In its simplest form, the problem of classifying gait disorders is a problem of mapping a multivariate temporal pattern to the most likely known disorder. Robust analysis of these data requires consideration of interactions among a large number of highly coupled variables, and the time dependence of these variables. Two approaches have been utilized: statistical techniques and artificial intelligence techniques.

Statistical Techniques

Several statistical techniques have been applied to the analysis of gait data. These include the "bootstrap" method (99,101), the linear discriminant method (102–104), principal component analysis (105), and cluster analysis (106). The *bootstrap technique* (107) was used to establish boundaries about the mean curve for unimpaired subjects (controls) to mark the limits of normal variability (99,101). These boundaries were designated as prediction regions. This technique was undertaken after initial attempts at setting boundaries for the variability within normal subjects using ensemble averages of one or two standard deviations failed. Kelly and Biden (108) compared the results of classification of knee motion by ensemble averaging versus bootstrapping. The motion curves of 39 unimpaired 5-year-old children were classified using both techniques. The ensemble-averaging method utilizing ± 2 standard deviations misclassified 16 of 39 normal subjects as abnormal. In contrast, the bootstrapping method classified all subjects as "normal." Bootstrap estimates of the prediction regions are of the form:

$$\hat{F}_h(\Theta) - m\hat{\sigma}_f(\Theta) \leq \tilde{F}_h(\Theta) \leq \hat{F}_h(\Theta) + m\hat{\sigma}_f(\Theta)$$

where $\hat{F}_h(\Theta)$ = the latest squares estimate of the subject's sum of harmonic coefficients

$\hat{\sigma}_f(\Theta)$ = the standard deviation of the harmonics, and

m = a positive number.

This technique has been applied clinically and has been shown to have a high sensitivity (109).

Methods of *discriminant analysis* have been shown to be effective in recognizing gait patterns of sound subjects and persons with gait deviation following total knee replacement surgery with a classification error rate of about 2 percent (102). This technique has also been used to develop knee and hip performance indices with well-demonstrated utility (103,104).

Principal component analysis and *cluster analysis* techniques have been used as a stepwise pattern-recognition approach to identify patterns of gait deviations. *Principal component analysis* is used to reduce the enormous quantity of data obtained in a gait study to a parsimonious set of features that describes gait patterns accurately (105) and results in a reduction in dimensionality of the original set of waveforms. Individual waveforms can be reconstructed using a linear combination of basis vectors modulated by weighting coefficients. Numerical representation using principal component analysis is important for two reasons (105): first, it is a parsimonious representation of cyclic subgroups within a larger patient population, and second, it may be very useful in identifying and classifying homogeneous subgroups within a larger patient population. *Cluster analysis* is used to place objects into groups or clusters suggested by the data, not defined *a priori*. Subjects in a given cluster tend to be similar to each other in some sense and subjects in different clusters tend to be dissimilar. These techniques have been used for classifying unimpaired subjects (110), persons with spastic paralysis (111), and persons with anterior cruciate ligament (ACL) deficiency (112).

Methods based on statistical analysis will continue to play a role in the processing of gait data. The strengths of statistical methods are that they provide a mathematical foundation for the analysis, accept experimental noise in the measurements, and offer robust time-series analysis. The weaknesses of statistical meth-

ods are that they ignore the physical meaning of the measurements and treat each variable in isolation.

Artificial Intelligence

An alternative approach to the analysis of gait dynamics is to use artificial intelligence (AI) techniques to diagnose gait disorders. Two categories of AI that have been used successfully are knowledge-based systems and neural networks.

Knowledge-based systems are most commonly referred to as "expert systems" and are characterized by large amounts of domain-specific knowledge and methods that embody the clinician's problem-solving strategy (113–115). Expert systems organize a knowledge base of facts that can be used to explain the logical connection between gait parameters and gait functional deficits. The facts in the knowledge base are arranged in premise-conclusion pairs called rules. The rules serve the purpose of causally relating gait parameters and functional deficits. The rules are probabilistic in nature, so inferences made by the program are seldom "all or nothing." The strength of expert systems is that they encapsulate high-level knowledge from "experts," and they model interactions among variables. However, the drawback of expert systems is that they assume that abnormal gait has been classified, and they only weakly model time.

A second method of AI is the *neural network*. Neural network designs are based on the structure of the human brain and try to emulate the way intelligent information processing occurs. The basic structure of a neural network is very simple. It consists of an array of elements usually called nodes, interconnections between these nodes, and some input/output scheme (**Figure 15**). The intelligent information properties of the network arise from the formation of the topology of the network, the learning rules of the nodes, and the particular type of nodes. Neural networks, despite their simplification of natural behavior, process information in novel ways. These networks have collective computational properties, such as association, generalization, differentiation, preferential learning, optimization, and fault tolerance. The use of these properties appears to have promise for the development of solutions to problems that have intractable or unknown algorithms and/or are too computationally intense. Neural networks follow an adaptive information-processing method well-suited for modeling dynamic processes.

Neural networks, which are capable of performing pattern-recognition tasks useful in the analysis of gait dynamics (116,117), have been shown to be capable of performing difficult temporal pattern processing tasks of gait kinematic data (118). The specific type of neural network used was a modification of standard back propagation as described by Elman (119). The network consisted of 12 input units, 10 hidden units, and 12 output units (**Figure 15**). It was trained using a set of 25 simulated unimpaired 7-year-old individuals. The simulated individuals were generated from the mean and variance data for the "normal" population (99). For each time increment, 12 motion variables, which contained the sagittal, coronal, and transverse plane motions of the pelvis, hip, knee, and ankle, were input to the neural network. The output of the network was the 12-tuple of motion variables at time increment, t+1. In this way, the network was trained to learn the temporal pattern of gait motion. The data set was subdivided into 50 time steps of each variable over a single gait cycle. After training was completed, the neural network was presented with gait patterns for 25 children at each age increment from 1 to 7 years of age. The difference between the new gait pattern (y′) and the learned gait pattern (y) was analyzed where the output error was calculated as follows:

$$SS_{Error} = \Sigma_i \Sigma_j (y - y')^2$$

where i = number of individual data sets (25) at each age increment, and
j = number of gait cycle divisions (50).

This total sum-squared error measures the deviation of each age group from 7-year-old gait (**Figure 16**). The results provide evidence that gait stabilizes between the ages of 3.5 and 4.0 years. This characteristic of gait development is supported by both expert physicians (99) and previous statistical analysis (101). This example demonstrates that neural networks are capable of performing pattern-recognition techniques useful in the analysis of gait dynamics.

In the future, neural networks can be used to differentiate normal and pathological gait. A person's gait data will be analyzed to yield a total sum-squared error. If the value exceeds a threshold, the person's gait will be further analyzed to pinpoint the areas of gait deviations, based on the difference between the individual's gait pattern and the learned normal gait pattern. Additional networks can be developed to differentiate subcategories of gait abnormalities. Once the individual's gait has been identified as abnormal, it can be analyzed by subsequent neural networks that are trained to recognize predefined functional gait deficits. Thus, it

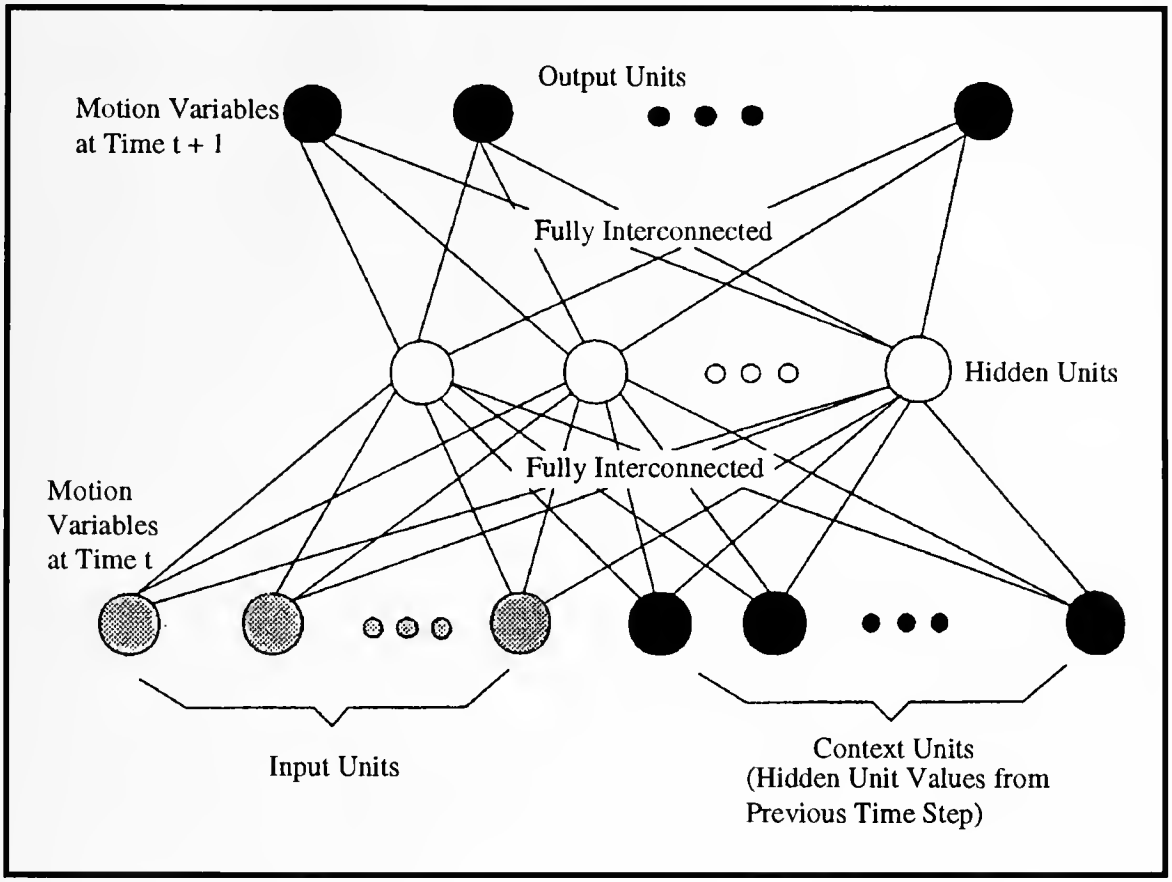

Figure 15.
Basic structure of a neural network (consisting of elements called nodes, interconnections between nodes, and an input/output scheme). This particular network is called a back propagation network and has a set of hidden nodes. It was used for temporal-pattern processing of gait kinematic data, which consisted of 12 motion parameters (118).

will be possible to identify the gait abnormalities of a subject.

Similar to the other techniques for classification of gait data, the strengths and weaknesses of neural networks must be recognized. The advantages of neural networks are that they capture the temporal structure of the gait variables, model the interconnection among these variables, and contain nonlinear processing elements. These advantages must be weighed against several disadvantages. First, neural networks require a large amount of data on which to be trained. Aside from unimpaired subjects, this amount of data on select pathologies might not be available. Further, neural networks require extensive training time in order to assure stable operation. Finally, neural networks do not distinguish between signal and noise (120).

Experienced specialists are needed to ensure that techniques used for pathological gait classification are reasonable. Each of the methods (statistical techniques, expert systems, and neural networks) offers advantages and disadvantages. The relative merits of each approach have not been fully investigated. In the end, it will be important to draw upon the strengths of all techniques in a productive and mutually supportive relationship in order to maximize the outcome.

TELECOMMUNICATION

Distributed data and computing resources need to be incorporated in the scientific computing environment of every motion analysis laboratory. Users must be able to gain transparent access to data and computing resources located anywhere in the world. Clinicians and researchers scattered around the globe should be connected via a network. Individuals operating computer software environments residing on their desktops should be able to communicate with leading centers in gait analysis.

Once standards have been established, it will be possible to share information among medical centers in order to obtain additional expert opinions on difficult cases. Currently, efforts are being made to create a national information infrastructure—the so-called electronic superhighway. This electronic network will carry voice, data, and video in digital form. At present, an electronic network, the Internet, already exists. The Internet was established in 1969 as an experimental computer network organized and financed by the Department of Defense and the National Science Foundation. The network was created to facilitate the research of a small number of scientists, engineers, and researchers. No commercial usage was permitted at first. Over time, the number of users of the Internet has increased. Currently, it is estimated that there are over 15 million users (121). Most of these users are in the United States but there are users in 134 other countries as well (121). The number of commercial users is also increasing. In early 1993, more than half of the registered networks were private businesses (121).

A nationwide communication system can be used in health care. High-performance computing and networking can be used to speed development of gait interpretation techniques, facilitate diagnoses from remote locations, and achieve enormous improvements in efficiency by aiding multicenter studies on treatment techniques. Major medical centers are obtaining state-of-the-art telecommunication capabilities (122); this includes the transmission of data, audio, and visual information. Telecommunication networks have made people throughout the world accessible within a matter of minutes or hours. It is no longer necessary for collaborators to be near one another. Current telecommunication systems provide two-way video and two-way audio. The image must be high-resolution and obtained in real time so that medical examinations can be performed. This connection will enable the transmission of information over high-bandwidth networks for immediate physician-physician consultation on particular cases. The security of all transmissions must be assured by scrambling the signal to maintain the confidentiality of all patient information.

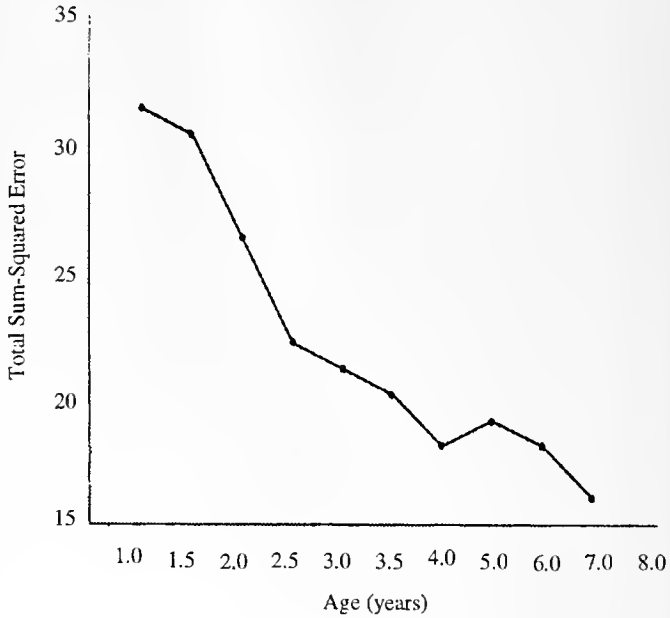

Figure 16.
Deviations of age group kinematic data from normal 7-year-old gait. Deviations expressed as a sum-squared error. Differences determined using a back propagation neural network. Results provide evidence that gait kinematics stabilize between 3.5 and 4.0 years of age (118).

In the years ahead, fiber-optic transmission and high-definition television will be among the advancements that will strengthen the interchange of information. Telecommunication will also enable the sharing of digital data with large bandwidth requirements for research purposes. The ability to share information will facilitate the development of databases that will enable clinicians to obtain knowledge for the treatment of specific gait disorders.

A national database for motion analysis data must be created. This national data repository can be used to pool gait assessments from participating centers throughout the United States. The database system should be based upon commercially available database software that can be used to maintain the data repository. Interaction with the database from participating centers would be through a web interface. The transmission of data would be protected through the secure sockets layer (SSL) protocol. This provides both en-

cryption of the data and authentication of each participating center. An Internet accessible web server that supports the SSL protocol should be used. This web server would offer hypertext pages, software applications and connectivity to the database server. Access to this web server would be permitted only for participating centers.

Each center would collect information on patients who have undergone biomechanical, neurological, and radiological evaluation relevant to clinical treatment for neuromuscular disorders (**Figure 17**). The data would be reduced to a standard data format, through the use of an application existing on each center's local workstation, and submitted to the repository website over the Internet (**Figure 18**) using an encrypted file. An online submission form would be completed. Subsequently, a completed report file would be returned to the center via electronic mail upon a successful load of the data into the data repository. Sensitive information, such as identifying information related to patients and surgeons, would be identified as such upon submission and recoded to ensure confidentiality. Similarly, each center would be able to extract information for its own use, including data submitted by other centers. In this case, an interactive form would assist in the search criteria used to extract the data sets, which would then be bundled together as a set of standard format files and made available for transfer from the multi-center web server in a location accessible only to each individual center. Notification of the completion of the query would be via electronic mail, which would include a retrieval summary of the resulting data set. The bundled data set selected by the center would be encrypted and transferred via the Internet to the center's workstation. This data could then be used to make outcome comparisons of patients who had been treated for similar physical conditions.

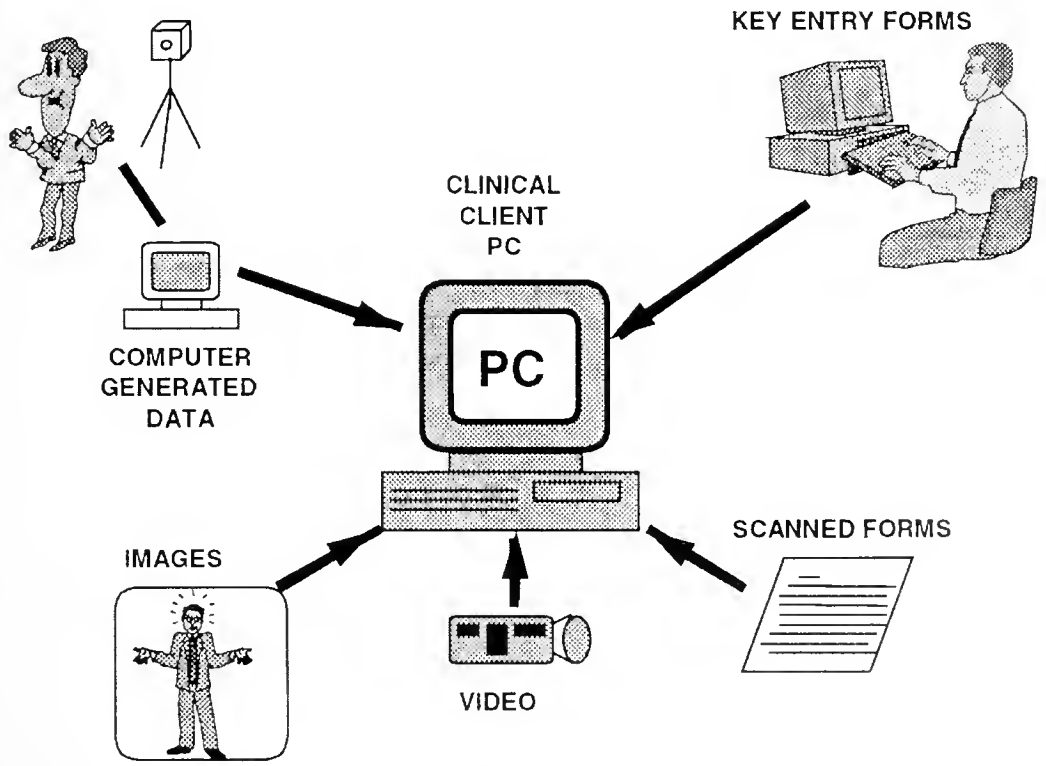

Figure 17.
Multimedia entry of patient data into a database.

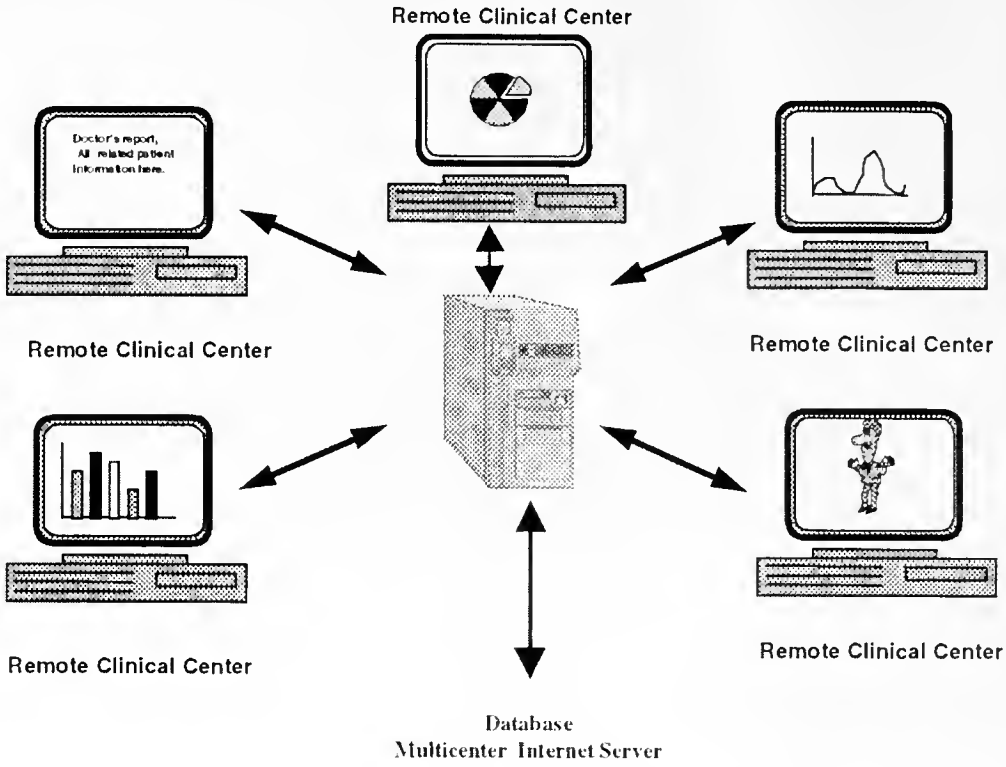

Figure 18. Remote storage/retrieval gait analysis data into a database repository.

SUMMARY

The ultimate goal of gait analysis is to provide reliable, objective data on which to base clinical decisions. A gait analysis laboratory requires an interdisciplinary team of individuals with various educational backgrounds who contribute their skills and who need to understand the underlying principles utilized to identify and correct neuromuscular deficiencies. The computer revolution will aid in developing new paradigms for computerized human movement analysis. New experimental techniques will be developed that will allow us to obtain real-time motion measurements. Computer animation techniques will become available to visualize gait data in an intuitive manner. Improvements will be made in our ability to obtain *in vivo* measurements of muscle function. Advances in both forward and inverse biomechanical models will continue. The future of gait analysis will require the ability to identify the critical tests, interpret data more quickly, predict the outcome of various clinical procedures, and quantify the outcome. Gait classification techniques will allow this to happen. Regional and national computer telecommunication networks need to be established whereby data can be exchanged to assimilate the knowledge necessary to predict the outcome of various surgical procedures. Efforts are underway to standardize techniques in order to facilitate the exchange of data. Reforms in health care will require that we be able to manage costs while providing an important clinical service. It is increasingly important that we consider the effectiveness of gait analysis and the role it will play in shaping the outcome of medical care.

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REFERENCES

1. Braune W, Fischer O. Der Gang des Menschen. Leipzig: BG Tenbner; 1895.

2. National Center for Health Statistics. Health, United States, 1995. Hyattsville, MD: Public Health Service; 1996.
3. Lamm, RD. The ghost of health care future. Inquiry 31(4):365-367, winter 1994/1995.
4. Gade C. Strategic Planning: Mayo faces changing times with an unchanged focus: the patient. Mayo Alumni 1993; 29(2):4–13.
5. Managed care raises quality and lowers hospital costs. Managed Care. Thousand Oaks, CA: Cauman Publications; August/September 1995. p.3.
6. National Center for Health Statistics. Health, United States, 1994. Hyattsville, MD: Public Health Service; 1995.
7. Group Health Association of America. HMO industry profile, Health Care Advisory Board analysis. 1994.
8. DeLuca PA, Ounpuu S, Rose SA, Sirkin R. Alterations in cerebral palsy surgical decision-making based on three-dimensional gait analysis. Dev Med Child Neurol 1993;35(9) suppl 69:9.
9. Nene AV, Evans GA, Patrick JH. Simultaneous multiple operations for spastic diplegia. J Bone Joint Surg Br 1993; 75-B(3):488–94.
10. Robinson L. Encyclopedia Americana 7:473. Danbury, CT: Grolier; 1988.
11. Murray MP, Drought AB, Kory RC. Walking patterns of normal men. J Bone Joint Surg 1964;46A:335–60.
12. Sutherland DH, Hagy JL. Measurement of gait movements from motion picture films. J Bone Joint Surg 1972;54A:787.
13. Chao E-YS. Justification of triaxal goniometry for the measurement of joint rotation. J Biomech 1980;15:989–1006.
14. Lamoreux LA. Kinematic measurements in the study of human walking. Bull Prosthet Res 1971;3:10-15:3–84.
15. Winter DA, Sidwall HG, Hobson DA. Television-computer analysis of kinematies of human gait. Comp Biomed Res 1972;5:498–504.
16. Cappozzo A. Gait analysis methodology. Hum Mov Sci 1984;3:27–50.
17. Antonsson EK, Mann RW. Automatie 3-D gait analysis using a Selspot centered system. Advances in bioengineering. New York: American Society of Mechanical Engineers; 1979. p. 51.
18. Andriacchi TP, Hampton SJ, Schultz AB, Gelante JO. Three-dimensional coordinate data processing in human motion analysis. J Biomech Eng 1979;101:279–83.
19. Elgie H. What is Scientific Visualization? Sci Comput Auto March, 1993;34–5.
20. Morris T, Larson G, Donath M. Real time animation of human walking for the evaluation of pathological gait. Proceedings of the 9th Annual RESNA Conference, Minneapolis, MN. 1986;233–5.
21. Tufte ER. The visual display of quantitative information. Cheshire, CT: Graphics Press; 1983.
22. Perry J, Waters RL, Perrin T. Electromyographic analysis of equinovarus following stroke. Clin Orthop 1978;131:47–53.
23. Sutherland DH, Cooper L, Daniel D. The role of the ankle plantar flexors in normal walking. J Bone Joint Surg Am 1980;62A:354–63.
24. Waters RL, Frazier J, Garland DE. Electromyographic gait analysis before and after operative treatment for hemiplegic equinus and equinovarus deformity. J Bone Joint Surg Am 1982;64A:284–8.
25. Blix M. Die langrund die spennung des muskels. Skand Arch Physiol 1894;5:149–206.
26. Cavanagh PR, Komi PU. Electromechanical delay in human skeletal muscle under concentric and eccentric contractions. Europ J Appl Physiol 1979;42:159–63.
27. Long C. Normal and abnormal motor control in the upper extremities (thesis). Cleveland, OH: Case Western Reserve University; 1970. p. 8.
28. Ralston HJ, Todd FN, Inman VT. Comparison of electrical activity and duration of tension in the human rectus femoris muscle. Electro Clin Neurophys 1976;16:277–86.
29. Norman RW, Komi PV. Electromechanical delay in skeletal muscle under normal movement conditions. Acta Physiol Scand 1979;106:241–8.
30. Vos EJ, Mullender MG, van Ingen Schenau GJ. Electromechanical delay in the vastus lateralis muscle during dynamic isometric contraction. Europ J Appl Physiol 1990;60:467–71.
31. Hill AV. The pressure developed in muscle during contraction. J Physiol 1948;107:518–26.
32. Kirkebö A, Wisnes A. Variation in tissure fluid pressure in rat calf muscle during sustained contraction on stretch. Acta Physiol Scand 1982;114:551–6.
33. Sutherland DH, Woo SLY, Schoon J, Jemmott G, Akeson WH. The potential application of a small solid state pressure transducer to measure muscle activity during gait. Trans Orthop Res Soc 1977;2:289.
34. Hargans AR, Sejersted OM, Kardel KR, Bloom P, Harmansen L. Intramuscular fluid pressure: a funetion of contraction force and tissue depth. Trans Orthop Res Soc 1982;7:371.
35. Mubarak S, Hargans A, Owen C, Garetto L, Akeson W. The Wick catheter technique for measurement of intramuscular pressure. J Bone Joint Surg Am 1976;58A:1011–9.
36. Owen CA, Garetto LP, Hargens AR, Schmidt DA, Mubarak SJ, Akeson WH. Relationship of intramuscular pressure to strengthen muscular contraction. Trans Orthop Res Soe 1977;2:246.
37. Parker PA, Körner L, Kadefors R. Estimation of muscle force from intramuscular total pressure. Med Bio Eng Comp 1984;22:453–7.
38. Körner L, Parker P, Almström C, et al. Relationship of intramuscular pressure to the force output and myoeclectric signal of skeletal muscle. J Orthop Res 1984;2:289–96.
39. Järvholm U, Palmerud G, Karlsson D, Herbertz P, Kadefors R. Intramuscular pressure and electromyography in four shoulder muscles. J Orthop Res 1991;9:609–19.
40. Garfin SR, Tipton CM, Mubarak SJ, Woo SLY, Hargens AR, Akeson WH. Role of fascia in maintenance of muscle tension and pressure. J Appl Physiol 1981;51:317–20.
41. Baumann JU, Sutherland DH, Hänggi A. Intramuscular pressure during walking: an experimental study using the Wick catheter technique. Clin Orthop 1979;145:292–9.
42. Kaufman KR, Sutherland DH. Dynamic intramuscular pressure measurement during gait. Oper Tech Sports Med 1995; 3(4):250–5.
43. Davis RB, Ounpuu S, Tyburski D, Gage JR. A gait analysis data collection and reduction technique. Hum Mov Sci 1991;10(5):575–87.

44. Kadaba MP, Ramakrishnan HK, Wootten ME. Measurement of lower extremity kinematics during level walking. J Orthop Res 1990;8:383–92.
45. Apkarian J, Naumann S, Cairns B. A three-dimensional kinematic and dynamic model of the lower limb. J Biomech 1989;22(2):143–55.
46. Cappozzo A, Leo T, Pedotti A. A general computational method for the analysis of human locomotion. J Biomech 1975;8:307–20.
47. Kaufman KR, An KN, Chao EYS. A dynamic mathematical model of the knee joint applied to isokinetic exercise. In: Spilker RL, Simon BR, editors. Computational methods in bioengineering. New York, NY: ASME: Biomechanical Engineering Division; 1988:9. p. 157–67.
48. Chao EYS. Justification of tri-axial goniometer for the measurement of joint rotation. J Biomech 1980;13:989–1006.
49. Grood ES, Suntay WJ. A joint coordinate system for the clinical description of three-dimensional motions: application to the knee. J Biomech Eng 1983;105:136–44.
50. Kinzel GL, Hall AS, Hillberry BM. Measurement of the total motion between two body segments: Part 1— analytic development. J Biomech 1972;5:93–105.
51. Spoor CW, Veldpaus FE. Rigid-body motion calculated from spatial coordinates of markers. J Biomech 1980;13:391–3.
52. Woltring HJ, Huiskes R, DeLange A, Veldpaus FE. Finite centroid and helical-axis estimation from noisy landmark measurements in the study of human joint kinematics. J Biomech 1985;18(5):379–89.
53. Woltring HJ. Analytical body-segment photogrammetry. In: Leo T, editor. Models, connection with experimental apparatus and relevant DSP techniques for functional movement analysis. Ancona, Italy: Dipartimento di Electronica ed Automatica, Universita di Ancona; 1990.
54. Chao EYS. Determination of applied forces in linking systems with known displacements: with special application to biomechanics (dissertation). Iowa City: University of Iowa; 1971.
55. Robertson DGE, Winter DA. Mechanical energy generation, absorption and transfer amongst segments during walking. J Biomech 1980;13:845–54.
56. Cappozzo A, Berme N. Subject-specific segmental inertia parameter determination—a survey of current methods. In: Berme N and Cappozzo A, editors. Biomechanics of human movement: applications in Rehabilitation, sports and ergonomics. Worthington, OH: Bertec Corp;1990. p. 179–85.
57. Dempster WT. Space requirements of the seated operator. WADC Technical Report 55-159. Wright-Patterson AFB; OH: AERO Medical Laboratory; 1955.
58. Clauser CE, McConville JT, Young JW. Weight, volume, and center of mass of segments of the human body. AMRL-TR-69-70 (AD 710 622). Wright-Patterson AFB, OH: Aerospace Medical Research Laboratory; 1969.
59. Chandler RF, Clauser CE, McConville JR, Reynolds HM, Young JW. Investigation of inertial properties of the human body. DOT HS-801. Washington, DC: National Highway Traffic Safety Administration; 1975. p. 430.
60. Drillis RJ, Contini R. Body segment parameters. Technical Report No. 1166.03 School of Engineering and Science, New York University; 1966.
61. McConville JT, Churchill TD, Calepis I, Clauser CE, Cuzzi J. Anthropometric relationships of body and body segment moments of inertia. Technical Report No. AFAMRL-TR-80-119. Wright-Patterson AFB; OH: Aerospace Medical Research Laboratory; 1980.
62. Young JW, Chandler RF, Snow CC, Robinette KM, Zehner GF, Lofber MS. Anthropometric and mass distribution characteristics of the adult female. Technical Report No. FAA-AM-83-16. Oklahoma City, OK: FAA Civil Aeromedical Institute; 1983.
63. Yeadon RM, Morlock M. The appropriate use of regression equations for the estimation of segmental inertial parameters. J Biomech 1989;22:683–9.
64. Hanavan EP. A mathematical model for the human body, Report No. AMRL-TR-102. Wright-Patterson AFB; OH: Aerospace Medical Research Laboratory; 1964.
65. Jensen RK. Estimation of the biomechanical properties of three body types using a photometric method. J Biomech 1978;11:349–58.
66. Jensen RK. Body segment mass, radius, and radius of gyration proportions of children. J Biomech 1986;19:359–68.
67. Hatze, HA. A mathematical model for the computational determination of parameter values of anthropometric segments. J Biomech 1980;13:833–43.
68. Sarfaty O, Ladin Z. A video-based system for the estimation of the inertial properties of body segments J Biomech 1993;26(8):1011–6.
69. Martin PE, Mungiole M, Marzke MW, Longhill LM. The use of magnetic resonance imaging for measuring segment inertial properties. J Biomech 1989;22:367–76.
70. Mungiole M, Martin PE. Estimating segment inertial properties: comparison of magnetic resonance imaging with existing methods. J Biomech 1990;23:1039–46.
71. Ashley S. Rapid prototyping for artificial body parts. Mech Eng 1983;50–3.
72. McMillan T. 3-D digitizing. Computer Graphics World January; 1989.
73. Chow CK, Jacobson DH. Studies of human locomotion via optimal programming. Math Biosci 1971;10:239–306.
74. Chow CK, Jacobson DH. Further studies of human locomotion: postural stability and control. Math Biosci 1972;15:93–108.
75. Townsend MA, Seireg A. The synthesis of bipedal locomotion. J Biomech 1972;5:71–83.
76. Mochon S, McMahon TA. Ballistic walking: an improved model. Math Biosci 1980;52:241–60.
77. Onyshko S, Winter DA. A mathematical model for the dynamics of human locomotion. J Biomech 1980; 13:361–8.
78. Hatze HA. Quantitative analysis, synthesis and optimization of human motion. Hum Mov Sci 1984;3:5–25.
79. Marshall RN, Jensen RK. A general Newtonian simulation of an N-segment open chain model. J Biomech 1985;18(5):359–67.
80. Nagurka ML. Theoretical approach for optimal motion generation of a bipedal locomotion model. Advances in Bioengineering, New York, NY: ASME;1986. p. 115–6.
81. Pandy MG, Berme N. A numerical method for simulating the dynamics of human walking. J Biomech 1988;21:1043–51.

82. Yamaguchi GT. Feasibility and conceptual design of functional neuromuscular stimulation systems for the restoration of natural gait to paraplegics based on dynamic musculoskeletal models (dissertation). Stanford, CA: Stanford University; 1989.
83. Meglan DA. Enhanced analysis of human locomotion (dissertation). Columbus, OH: Ohio State University; 1991.
84. Zajac FE. Muscle coordination of movement: a perspective. J Biomech 1993;26(1):109–24.
85. Yamaguchi GT, Pandy MG, Zajac FE. Dynamic musculoskeletal models of human locomotion: perspectives on model formulation and control. In: Patla A, editor. Adaptability of human gait: implications for the control of locomotion. Advances in Psychology Series No. 78. Amsterdam: Elsevier Science Publishers;1991. p. 205–40.
86. Pandy MG, Anderson FC, Hull DG. A parameter optimization approach for the optimal control of large-scale musculoskeletal systems. J Biomech Eng 1992;114:450–60.
87. Zajac FE. Muscle and tendon: properties, models, scaling, and application to biomechanics and motor control. In: Bourne JR, editor. CRC Critical Reviews and Biomedical Engineering Boca Raton, FL: CRC Press; 1989. 17(4):349–411.
88. Hoy MG, Zajac FE, Gordon ME. A musculoskeletal model of the human lower extremity: the effect of muscle, tendon, and moment arm on the moment-angle relationship of musculotendon actuators at the hip, knee, and ankle. J Biomech 1990;23:157–69.
89. Kaufman KR, An KN, Chao EYS. Incorporation of muscle architecture into the muscle length-tension relationship. J Biomech 1989;22(8/9):943–9.
90. Lieber RL, Brown CG, Trestik CL. Model of muscle-tendon interaction during frog semitendinosus fixed-end contractions. J Biomech 1992;25:421–8.
91. Trestik CL, Lieber RL. Relationship between Achilles tendon mechanical properties and gastrocnemius muscle function. J Biomech Eng 1993;115:225–30.
92. Brand RA, Crowninshield RD, Wittstock CE. A model for lower extremity muscular anatomy. J Biomech Eng 1982;104:304–10.
93. Delp SL, Loan JP, Hoy MG, Zajac FE, Topp EL, Rosen JM. An interactive graphics-based model of the lower extremity to study orthopedic surgical procedures. IEEE Trans Biomed Eng 1990;37(8):757–67.
94. Yamaguchi GT, Zajac FE. A planar model of the knee joint to characterize the knee extensor mechanism. J Biomech 1989;22:1–10.
95. Johnston RC, Brand RA, Crowninshield RD. Reconstruction of the hip: a mathematical approach to determine optimum geometric relationships. J Bone Joint Surg Am 1979;61A: 639–52.
96. Dul J, Shiavi R, Green N. Simulation of tendon transfer surgery. Eng Med 1985;14:31–8.
97. Lindgren U, Seireg A. Influence of mediolateral deformity, tibial torsion, and foot position on femoral tibial load: prediction of a musculoskeletal computer model. Arch Orthop Trauma Surg 1989;108:22–6.
98. Mann RW. Computer-aided surgery. Proceedings of the 8th annual RESNA conference, Memphis, TN; 1985: 26–35.
99. Sutherland DH, Olshen RA, Biden EN, Wyatt MP. The development of mature walking, Bax M, editor. London: MacKeith Press; 1988. p. 28–9.
100. Thomas SS. The gait analysis laboratory: an administrative manual for physicians and administrators. Results of master's thesis. Proceedings of the 7th Annual East Coast Clinical Gait Laboratory Conference, Richmond, VA, Oct 31-Nov 2, 1991.
101. Olshen RA, Biden EN, Wyatt MP, Sutherland DH. Gait analysis and the Boot Strap. Annal Stat 1989;17(4):1419–40.
102. Donath M. Human gait pattern recognition for evaluation, diagnosis and control (dissertation). Cambridge, MA: Massachusetts Institute of Technology; 1978.
103. Laughman RK, Stauffer RN, Ilstrup DM, Chao EYS. Functional evaluation of total knee replacement. J Orthop Res 1984;2:307–13.
104. Kaufman KR, Chao EYS, Callahan TD, Askew LJ, Bleimeyer RR. Development of a functional performance index for quantitative gait analysis. Biomed Sci Instrum 1987;23:49–55.
105. Wootten ME, Kadaba MP, Cochran GVB. Dynamic electromyography I: numerical representation using principal component analysis. J Orthop Res 1990;8:247–58.
106. Kadaba MP, Ramakrishnan HK, Jacobs D, Goode B, Scarborough N. Relationships between patterns of knee and ankle motion in spastic diplegic patients with dynamic ankle equinus. Trans Orthop Res Soc 1993;18(2):364.
107. Efron B. The Jack Knife, the Bootstrap, and other resampling plans. Philadelphia, PA: Society for Industrial and Applied Math; 1982.
108. Kelly MF, Biden EN. A comparison of two classification methods for gait data. Trans Orthop Res Soc 1989;14:241.
109. Sutherland DH, Kaufman K, Ramm K, Ambrosini D, Wyatt M. Clinical use of prediction regions for motion data. Dev Med Child Neurol 1996;38(9):773–81.
110. Wootten ME, Kadaba MP, Cochran GVB. Dynamic electromyography II: normal patterns during gait. J Orthop Res 1990;8:259–65.
111. Wong MA, Simon S, Olshen R. Statistical analysis of gait patterns of persons with cerebral palsy. Stat Meth 1983;2:345–54.
112. Shiavi R, Zhang LQ, Limbird T, Edmondstone MA. Pattern analysis of electromyographic linear envelopes exhibited by subjects with uninjured and injured knees during free and fast speed walking. J Orthop Res 1992;10:226–36.
113. Dzierzanowski JM, Bourne JR, Shiavi R, Sandell HSH, Guy D. Gaitspert: an expert system for the evaluation of abnormal human locomotion arising from stroke. IEEE Trans Biomed Eng 1985;32(11):935–42.
114. Simon SR, Bylander T, Weintraub M, Szolovits P, Hirsch DE. Doctor gait: an expert system for gait analysis. Trans Orthop Res 1989;14:245.
115. Hirsch DE, Simon SR, Bylander T, Weintraub MA, Szolovits P. Using causal reasoning in gait analysis. In: Horne W, editor. Causal AI models: steps toward applications. New York, NY: Hemisphere Publications; 1989. p. 253–72.
116. Holzreiter SH, Kohle ME. Assessment of gait patterns using neural networks, J Biomech 1993;26(6):645–51.
117. Sepulveda F, Wells DM, Vaughan CL. A neural network representation of electromyography and joint dynamics in human gait. J Biomech 1993;26(2):101–9.

118. Biafore S, Cottrell G, Focht L, Kaufman K, Wyatt M, Sutherland DH. Neural network analysis of gait dynamics. Trans Orthop Res Soc 1991;16(1):255.
119. Elman JL. Finding structure in time. Center for Research in Language, Tech Report No. 8801. San Diego: University of California; 1988.
120. Sartori DE. Neural networks, statistics and experimental designs. Sci Comput Auto 1992;8(11):4–6.
121. Schiller HI. Electronic highway to where? National Forum 1994;74(2):19–21.
122. Parker BK. Telecommunication: when the sky's the limit. Mayo Mag 1990;5(1):18–29.

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